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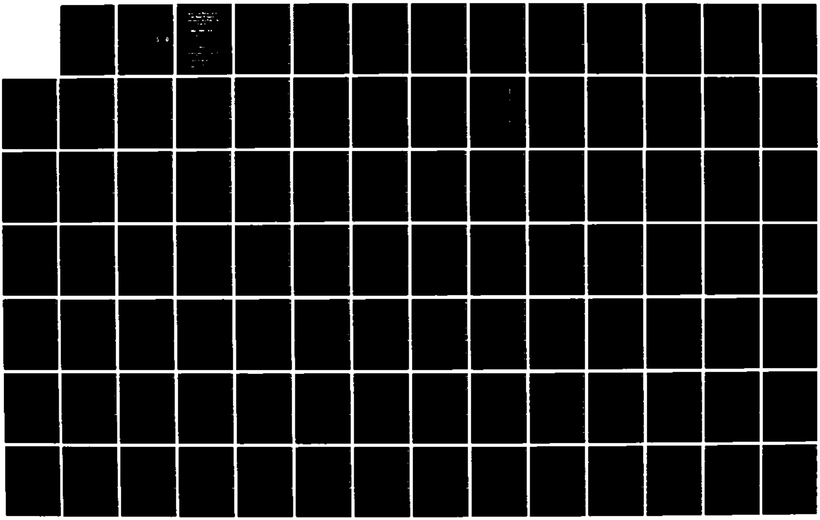

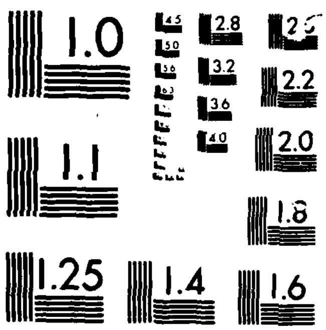

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Coordination Mechanism In Fast Human Movements - Experimental And Modelling Studies

VOLUME I

ANNUAL SUMMARY REPORT

Walter Kroll
William L. Kilmer

SEPTEMBER 1983

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20. ABSTRACT (Continue on reverse side if necessary and identify by block number)

Results of Year 3 are presented and includes: (a) a detailed report on the quantitative analysis of practice effects upon the triphasic EMG pattern for a maximum speed forearm flexion movement; (b) patterned electrical stimulation effects upon neuromotor coordination mechanisms underlying speed of forearm flexion movement speed; and (c) high frequency Russian type electrical stimulation of agonist and antagonist muscle groups involved in fast forearm flexion movement.

INTRODUCTION

The present study is investigating the basic neuromotor coordination mechanisms involved in a rapid elbow flexion movement, and in developing mathematical models to explain the interaction of these basic neuromotor coordinations with the biomechanical parameters of movement speed. Speed of movement is being assessed in a biomechanics mode via displacement, velocity, acceleration, point of inflection between acceleration and deceleration, and the total time of an elbow flexion movement. Electromyographic analysis techniques are used to monitor the sequential timing of agonist and antagonist muscle activity. The mathematical modelling effort incorporates the biomechanical parameters into an interface with the neurophysiological parameters involving the central and peripheral nervous systems, and then extends the interface to include viscoelastic properties of the muscle, activation delays, and neuronal pools.

([illegible] rate reactions; forearm flexion movement).

The experimental approach includes consideration of the neuromotor coordination mechanisms in both loaded and unloaded elbow flexion movements; changes in control mechanisms due to practice and learning effects; changes in control mechanisms due to local muscular fatigue induced by isometric exercise in the agonist and in the antagonist muscle groups; feasible training and practice regimens involving artificial means of enhancing beneficial changes in control mechanisms; and development of suitable mathematical models to explain in functional terms the ways in which the adaptive mechanisms can account for changes in basic coordination skill and the breakdown of skilled movement patterns due to local muscular fatigue. The planned series of studies incorporates research protocols from at least three usually distinct and isolated areas of research:

neurophysiology, biomechanics, and computer science and mathematical modelling.

MODELLING STUDIES

In the first year and a half we developed a successful EMG-level model for a fast arm movement to target which was published in Biological Cybernetics (1). Over the next year we used a simpler version of the EMG-based model to study feedback controls of elbow posture that compensate for torque-load perturbations. We found that when delays in the feedback loop are taken into account, an angular-position-servo control is always unstable and leads to spastic oscillations, whereas a mixed angular-velocity and angular-position servo control is stable at sufficiently low loop gains (control power). As described in the Journal of Mathematical Biology (2), pure angular-velocity servo control is the most stable arrangement of all. These results correlate well with experimental tests (12).

From these modelling efforts we hypothesized that unexpected step-torque disturbances to the elbow are usually compensated for by, first, bringing the elbow angular velocity to zero with an angular-velocity servo while at the same time estimating the new load torque, and then, second, returning the elbow to its original posture with an "optimal" volitional control. The optimality criterion to be minimized is a weighted blend of the time integral over the elbow movement of: (angular position error)^2, (angular velocity)^2, (angular acceleration)^2, (EMG-signal envelope)^2, (degree of flexor and extensor EMG co-activation), and (terminal error)^2. All EMG terms are with respect to both flexor and extensor. We have a computer algorithm (A) working which calculates the optimal volitional response to any postural disturbance for the nonlinear dynamic model

of (1), actually only a good nonlinear approximation to that model. This was difficult to achieve because of the continuous delays in a muscle's response to a nervous input (4). We solved the feedback control equation (6, 7) which now enables us to compute elbow angular responses under the hypothesis (H2): namely, that in the face of small, rapidly varying, zero-average perturbations in elbow-torque-load, the elbow angular response derives from an optimal control (above optimality criterion) based only on myotatic feedback from the arm. H2 follows naturally from the experimental work of Terzuolo et al. (11, 12). A third hypothesis (H3) which merits consideration is that (H2) is correct, but only with respect to a sampled date of mode of operation, with sampling intervals of about 500 ms (there is some evidence for this, cf. reference 11).

EXPERIMENTAL STUDIES

As a direct result of research completed during Years 1 and 2, the reverse loop theory of motor learning was formulated and presented at various conventions and invited talks. The purpose of these presentations was to elicit reactions and critiques of the theory for resolution in further studies. We have shown that improvement in limb movement speed can be produced by patterned electrical stimulation of relevant muscle synergy action patterns without any actual physical practice. Changes produced in the electromyographic patterns for limb movement tasks follow the patterned electrical stimulation protocols. Improvements in neuromotor coordination mechanisms and limb movement speed produced by the patterned electrical stimulation are presumed to be due to sensory imparted learning, and constitute experimental evidence for the reverse loop theory of motor learning. Both the reverse loop and sensory imparted learning theories

promise to alter radically present motor learning theories and result in an enhanced understanding of motor skill acquisition in general.

In order to validate properly the sensory imparted learning and reverse loop theories, it was planned to apply patterned electrical stimulation protocols based upon different combinations of the identified neuromotor coordination mechanisms described in our mathematical model of fast limb movements (Kilmer, Kroll, & Congdon, 1982). During Year 3 we proposed, therefore, a series of studies designed to elicit relevant experimental tests of the model. Briefly, the series of studies involved patterned electrical stimulation to produce changes in limb movement speed and neuromotor coordination mechanisms based upon different pattern protocols. Thus, if the pattern protocol alteration was in the agonist to antagonist latency parameter, the expectation would be for the agonist to antagonist latency parameter to be altered following administration of the patterned electrical stimulation sessions.

Results from the early series of studies also suggested that limb movement coordination mechanisms were markedly affected by inertial loading of the limb. In order to investigate this lead, we administered patterned electrical stimulation based upon neuromotor coordination mechanism parameters elicited under different inertial loads. The addition of the inertial load parameter necessitated additional studies over and above those originally planned for Year 3.

In order to gain further insight into the neuromotor coordination mechanisms a quantitative analysis of the raw electromyographic triphasic pattern was also undertaken. It was not known whether quantitative EMG analysis of muscle action potentials secured during a fast ballistic

movement would be reliable criterion measures. The need, however, to gain some information about the firing frequency and characteristics of motor unit firing pattern seemed to warrant the risk of collecting unreliable data. As it turned out, the quantitative EMG measures were reliable.

Finally, use was made of high frequency electrostimulators as a means of producing strength increases in specific muscle groups. The goal was to determine whether neuromotor coordination mechanisms can be altered by electrically produced strength increases in either the flexor or extensor muscle groups. If selective strength changes could be produced in the flexor or extensor muscle groups, the proposed mathematical model for neuromotor coordination mechanisms in fast ballistic movement could be tested. Traditional exercise regimens produce strength increases in both agonist and antagonist muscle groups because of co-contraction. Thus, if traditional exercise requires vigorous contraction of the elbow flexors, a strong antagonist elbow extension contraction is elicited due to co-contraction and strength improvement results in both agonist and antagonist muscle groups. Through use of the high frequency Russian type electrostimulator it was hoped that selective strength increases could be produced in either the agonist or antagonist muscle groups.

RESULTS

Appended to this summary of Year 3 research are detailed reports of each of the major investigative efforts. A summary of each of these three major efforts follows.

Quantitative Analysis of Practice Effects Upon the Triphasic EMG Pattern for a Maximum Speed Forearm Flexion Movement

Eight male and eight female subjects were tested for maximum speed forearm flexion movement under each of three different inertial load conditions on eight days. Daily alteration of two kinematic (movement time and percent acceleration time) and six raw EMG parameters were studied in detail. The six raw EMG parameters assessed were number of spikes (NOS), mean spike amplitude (MSA), mean number of peaks per spike (MNP), spike frequency (SF), mean spike duration (MSD), and mean spike slope (MSS). Four temporal components of the triphasic EMG pattern were also assessed: biceps brachii motor time (B1), end of the first biceps brachii burst (B2), second biceps brachii burst (B3), and triceps brachii burst (B4).

Results showed that all quantitative EMG parameters (except mean spike slope) were measured reliably for biceps brachii B1 and B2 components. Practice effects produced faster movement times and similar EMG spike parameter changes in biceps brachii B1 and B2 components. Practice effects produced a decrease in mean spike duration (MSD) of 12 and 13 percent for biceps brachii B1 and B2 components, respectively. Spike frequency (SF), on the other hand, increased eight and eleven percent for B1 and B2 components. Inertial loading was associated with an increase in the number of spikes (NOS) in B1 and B2 components. Movement time correlated $r = .57$ with number of spikes in the B2 component. Biceps brachii B2 component correlated negatively $r = -.65$ with percent acceleration time (PAT).

This investigation was the first to measure reliably quantitative EMG parameter changes due to practice effects in fast ballistic limb movement. The mean decrease of 15 milliseconds (10 percent) in forearm flexion movement time was associated with a shortening of the total biceps brachii burst, a 12 percent decrease in mean spike duration, and an eight to eleven percent increase in spike frequency. These results suggest that firing rate modulation is the major mechanism involved in motor unit firing pattern changes due to practice. Synchronization of motor unit firing or the recruitment of faster firing high threshold fast twitch motor units do not appear to be neuromotor coordination mechanisms involved in practice changes. Although higher threshold fast twitch motor units could produce a faster firing frequency, the reduction in mean spike duration negates such an explanation. Higher firing rates associated with fast twitch motor units would also produce longer spike durations which was not the case. Instead, the same motor unit pool is being recruited but made to fire at higher frequencies.

Patterned Electrical Stimulation Effects Upon Neuromuscular Coordination Control Mechanisms Underlying Speed of Forearm Flexion Movement

Thirty-six subjects were allocated equally to a control, a physical practice, and four different electrical stimulation experimental groups. Three pre-test stabilization days were followed by two treatment periods of two weeks duration with a post-test following each of the two week treatment periods. The four experimental groups were each administered

a different electrical stimulation protocol for six 30-minute sessions during each of the two week treatment periods. Each 30-minute session included an electrical stimulation protocol pattern being administered once every 10-seconds for a total of 180 patterns each session. Over six such sessions 1,080 electrical stimulation protocol patterns would be administered. In the second two-week treatment period another such 1,080 patterns would be administered.

The four electrical stimulation groups consisted of two high frequency and two low frequency groups. In both the high and the low frequency groups, a progression and a retrogression group also existed. Progression groups received a longer agonist to antagonist latency pattern while the retrogression groups received a shorter agonist to antagonist latency pattern. Since the agonist to antagonist latency parameter is one of the major neuromotor coordination mechanisms in fast limb movement it was hypothesized that manipulation of this pattern would produce modification in the triphasic EMG pattern as well as in task performance. The low frequency (Hz 50) stimulation is within the normal neurophysiologic range of motor unit firing while the high frequency stimulation (1,000 Hz) is not. The question of employing normal rather than abnormal wave form characteristics in electrical stimulation would thus be tested.

Following the initial baseline stabilization practice days, movement time was significantly affected by the patterned electrical stimulation pattern protocols. The progression (longer agonist to antagonist latency) groups were found to be slower after two-week treatment periods while the retrogression groups got faster. Antagonist muscle activity, particularly the IEMG slope parameter, was significantly affected by the

patterned electrical stimulation treatment periods. Except for a decrease in total EMG duration, agonist muscle activity parameters remained essentially unchanged. Prediction of movement speed was affected by practice effects. In the unpracticed condition, the agonist muscle activity was a more important predictor than the antagonist muscle. After extended practice, the antagonist muscle activity parameters were more important predictors.

Based upon neuromotor coordination mechanisms identified in previous studies, patterned electrical stimulation protocols were shown to be an effective technique with which to manipulate limb movement speed. Further, improvements in limb movement speed produced by patterned electrical stimulation were associated with predicted changes in the triphasic EMG pattern. Such changes in the triphasic EMG pattern were effectively manipulated by the patterned electrical stimulation protocols.

High Frequency Electrical Stimulation of Agonist and Antagonist Muscle Groups Involved in Fast Forearm Flexion Movement: Effects Upon Movement Time and the Triphasic EMG Pattern

Ten male and ten female subjects were allocated equally to two experimental groups. One group received electrical stimulation of the elbow flexion muscle group and the other group received electrical stimulation of the elbow extensor muscle group. Flexion and extension movement speed in unloaded and loaded conditions, EMG of biceps brachii and triceps brachii muscles during movement trials, maximum isometric strength, and endurance holding time with a load equal to 50 percent of maximum isometric strength were assessed over four baseline stabilization

days. Following baseline days, 18 sessions of electrical stimulation were administered using a high frequency (2500 Hz) Russian type stimulator. Two post-test sessions duplicated baseline stabilization day measurements.

The main conclusion reached was that high frequency electrical stimulation did not produce the strength increases reported by Russian investigators. Males did show a 23 percent increase in extension strength in the arm which received stimulation on the extensors. Only a 12.3 percent increase in extension strength occurred in males when the extensors were stimulated. One probable explanation for the above results in which extension strength increased more with flexor stimulation than with extension stimulation is that the stimulated flexor contraction was so intense it required subjects to co-contract the extensors to protect the joint. Such active co-contraction produced greater strength increases than actual electrical stimulation protocols.

Movement time changes occurred mostly in the male subjects. Males receiving flexor stimulation had approximately a 5 percent faster flexion movement time in both unloaded and loaded conditions. Flexor stimulation in males also produced 12.4 and 12.9 faster extension movement times in unloaded and loaded conditions, respectively. Males receiving extensor stimulation showed 12.4 percent faster extension movement time. Females displayed differential changes in movement speed due to electrical stimulation and no significant increases in strength.

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APPENDIX A

Quantitative Analysis of Practice Effects Upon the Triphasic EMG Pattern for a Maximum Speed Forearm Flexion Movement

PROCEDURES

Testing Schedule

Sixteen college age students, eight male and eight female, acted as subjects in the present study. Subjects reported to the Motor Integration Laboratory on eight separate practice days, the first three practice days at most 24 hours apart and the last five practice days at most 48 hours apart.

Each subject performed ten maximum speed forearm flexion trials (from 160 to 90 degrees; 180 degrees being full

extension) under three different loads (no load, four times, and eight times the forearm moment of inertia) on each practice day. At least 30 seconds rest was provided between each of the 30 trials performed every practice day to avoid fatigue. Every maximum speed forearm flexion trial was with the subject's right arm and was stopped by antagonist (triceps brachii) muscle action (i.e., class B movement). Data were recorded during each maximum speed forearm flexion trial executed on test days one, four, and eight. No data were recorded on practice days two, three, five, six, and seven, these days serving only as practice days.

Maximum Speed Forearm Flexion Quantification Procedures

As seen in Figure 5, every maximum speed forearm flexion trial was performed as a subject was seated on a stool with the chest strapped against a chest pad attached to a specially designed table. The stool height was adjusted so that the subject's right upper arm was positioned on the table, parallel to the floor, at a 90 degree angle to the trunk. A leather cuff strapped around the subject's wrist was attached to a wooden bar positioned parallel to the forearm. The bar was attached to a ball bearing joint which was fixed to the table and aligned to the center of rotation of the elbow joint. Each maximum speed forearm flexion trial was terminated when the subject's right

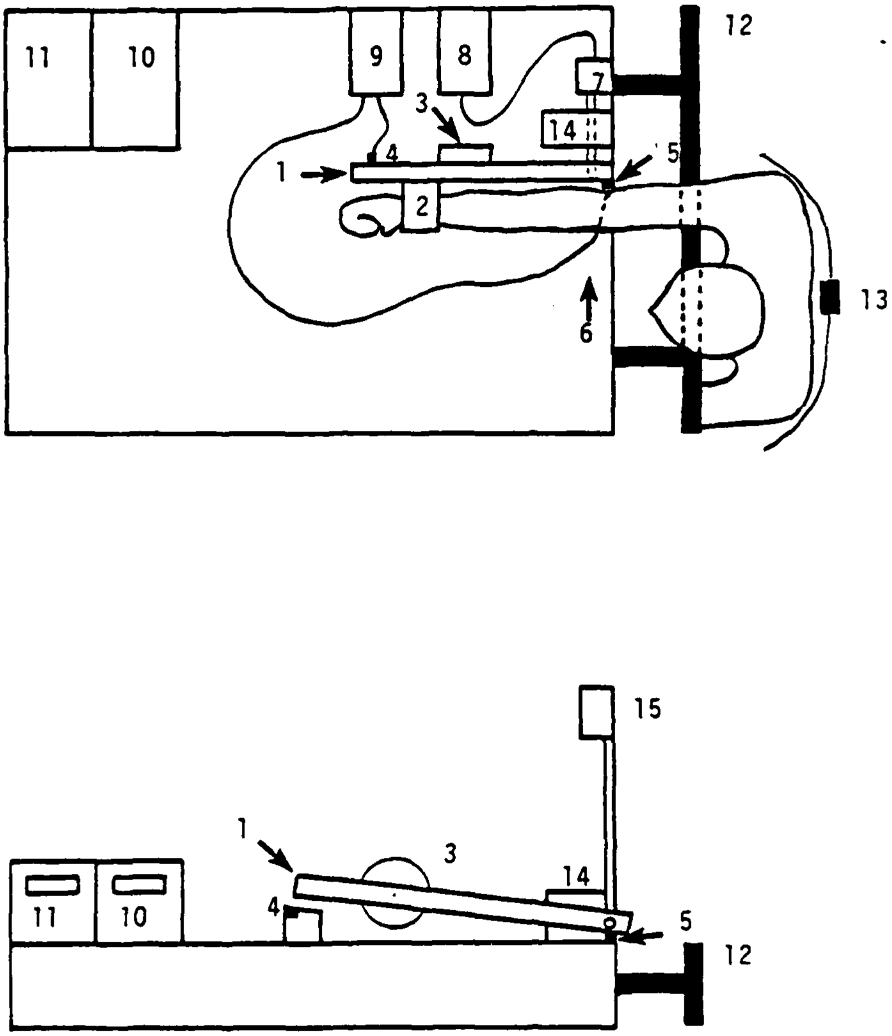


Figure 5. Movement apparatus. (1) wooden bar; (2) wrist cuff; (3) lead weight; (4) start microswitch; (5) end microswitch; (6) rotational axis; (7) potentiometer; (8) acceleration time integrated circuit; (9) movement time-event marker integrated circuit; (10) movement time digital clock counter; (11) acceleration time digital clock counter; (12) chest pad; (13) seat belt; (14) ball bearing joint; (15) target.

hand touched a target positioned perpendicular to the table passing through the rotational axis of the elbow joint as shown in Figure 5.

Three load conditions were determined for each subject specific to the moment of inertia of the subject's forearm, thus insuring that loading conditions were mechanically equal for all subjects. Load zero was assigned zero additional load while load one and load two were assigned four times and eight times the forearm moment of inertia respectively. The moment of inertia of each subject's forearm was estimated according to the procedures outlined by Plagenhoef [71] shown in Appendix B. Increments in imposed moment of inertia were accomplished by attaching a lead weight of constant known mass to the wooden bar adjacent to the subject's arm at a predetermined distance from the elbow's rotational axis. For all subject's a .45kg lead weight was used for load one while a .90kg lead weight was used for load two. Prior to testing on test day one, calculations of the distance from rotational axis to lead weight (i.e. radius of gyration; see Appendix B) for loads one and two were made and recorded so that the same distances could be used on each subsequent practice day.

Prior to each maximum speed forearm flexion trial, the wooden bar and forearm were placed in a ready position on a microswitch such that initiation of flexion activated the microswitch (see Figure 5). A second microswitch was activated upon completion of each maximal speed forearm flexion trial. In

conjunction with a specially designed movement time-event marker integrated circuit (designed by Everett Harman, Exercise Science Department, University of Massachusetts, Amherst), the microswithches served two main functions. First, they started and stopped a digital clock-counter (Lafayette Instrument Corp. model 54419) which timed maximum speed forearm flexion movement as shown in Figure 5. Second, they simultaneously activated movement start and movement end analogue event markers on a four channel Sony analogue recorder model TC-277-4 (frequency bandwidth of 30-200,000 Hz).

Attached to the ball bearing joint located at the center of rotation of the elbow joint was a potentiometer whose function in concert with another specially designed acceleration time integrated circuit [49] was to start and stop another digital clock-counter at the onset and end of acceleration of the forearm as shown in Figure 5. In this way, maximum speed forearm flexion acceleration time was quantified. Maximum speed forearm flexion acceleration time was expressed as a percentage of maximum speed forearm flexion movement time and denoted maximum speed forearm flexion percent acceleration time.

EMG Signal Quantification Procedures

Beckman Ag-AgCl bipolar surface electrodes were utilized to simultaneously pick up unfiltered analogue raw EMG signals from the long head of m. biceps brachii and the lateral head of m. triceps brachii during each maximum speed forearm flexion trial on test days one, four, and eight (see Figure 6). Active electrodes remained in place over each muscle's approximated motor point [10] only after the skin-electrode impedance was reduced to 5k ohms or less using standard skin preparation procedures [10]. Active electrodes were attached to the skin 4.25 cm apart center to center, in a position parallel to the muscle fiber direction. In addition, a ground electrode was attached to the skin overlying the right clavicle of each subject.

As shown in Figure 6, biceps and triceps brachii analogue raw EMG signals were differentially amplified and visually inspected during recording with a two channel Medic Flexline-S storage oscilloscope electromyograph model SNV2H4. The Medic electromyograph has a frequency bandwidth of 2 Hz to 20 KHz, an input impedance of 250 megohms with a common mode rejection ratio of 259,000 to 1. Analogue raw EMG signals from the biceps and triceps brachii muscles that passed visual inspection were subsequently stored with the previously discussed movement

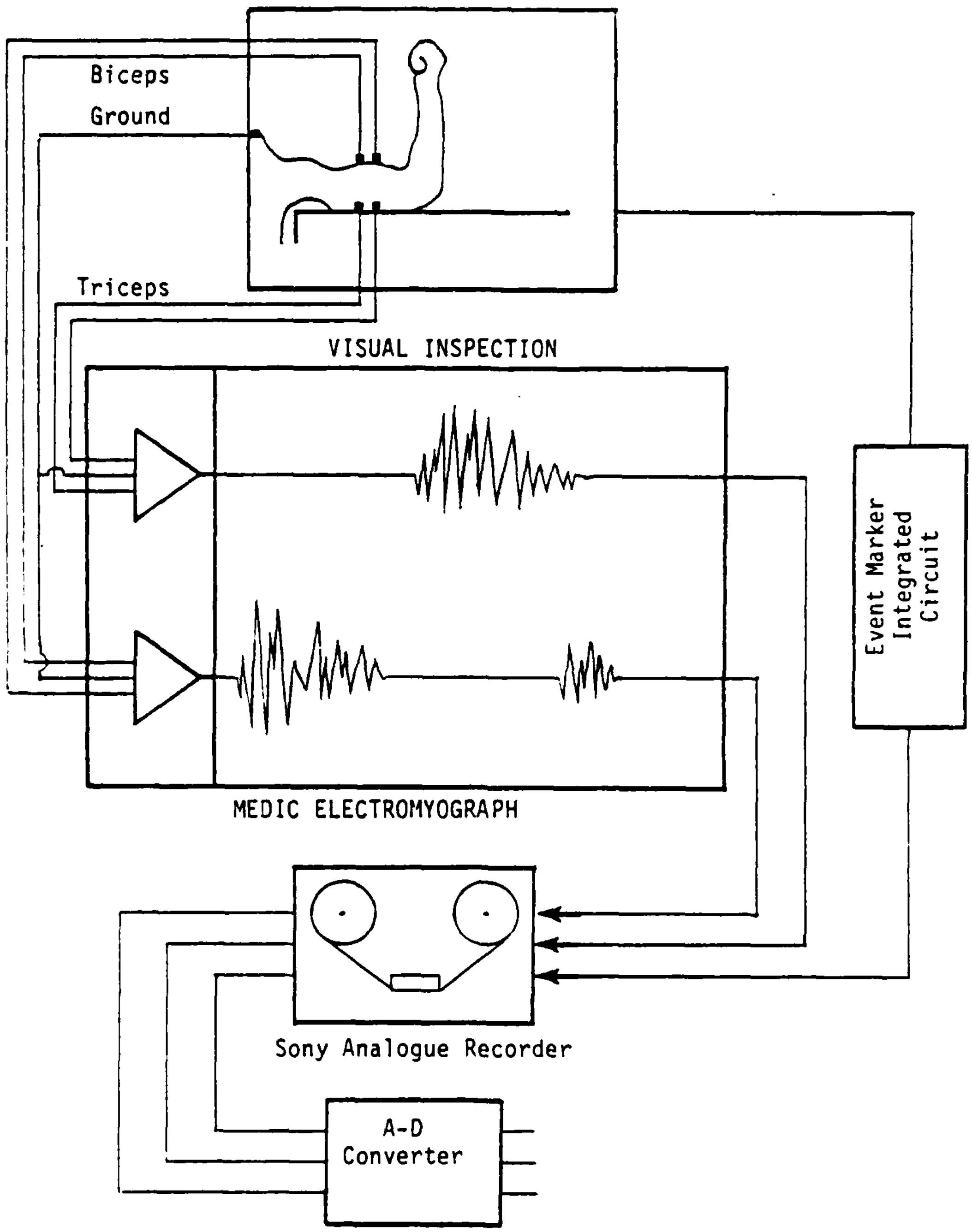


Figure 6. Analogue phase of EMG signal quantification

initiation and completion event markers on the Sony analogue recorder shown in Figure 6. The Sony analogue recorder had been previously calibrated so as to insure the reproducibility of every receded signal.

Analogue to digital conversion of biceps brachii analogue raw EMG signal and triceps brachii analogue raw EMG signals as well as the movement start and movement end analogue event markers was accomplished using an analogue to digital converter at a constant sampling interval of 2000 points per second (courtesy of Computerized Biomechanical Analysis Inc.). Thus, a second of biceps brachii analogue raw EMG signal, triceps brachii analogue raw EMG signal, and movement initiation and movement completion analogue event markers was converted into 2000 discrete digital x-y coordinates. Digitized biceps brachii raw EMG signals, triceps brachii raw EMG signals, and movement initiation and movement completion event markers were subsequently loaded onto hard disk within a Data General Micro-Nova Mini Computer.

As depicted in Figure 7, a user interactive graphics computer program, a Hewlett Packard CRT, and the Data General Micro Nova Mini Computer (all graciously provided by Computerized Biomechanical Analysis Inc.) enabled an "on line" visual selection of digital raw EMG signal duration during biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst. First, the

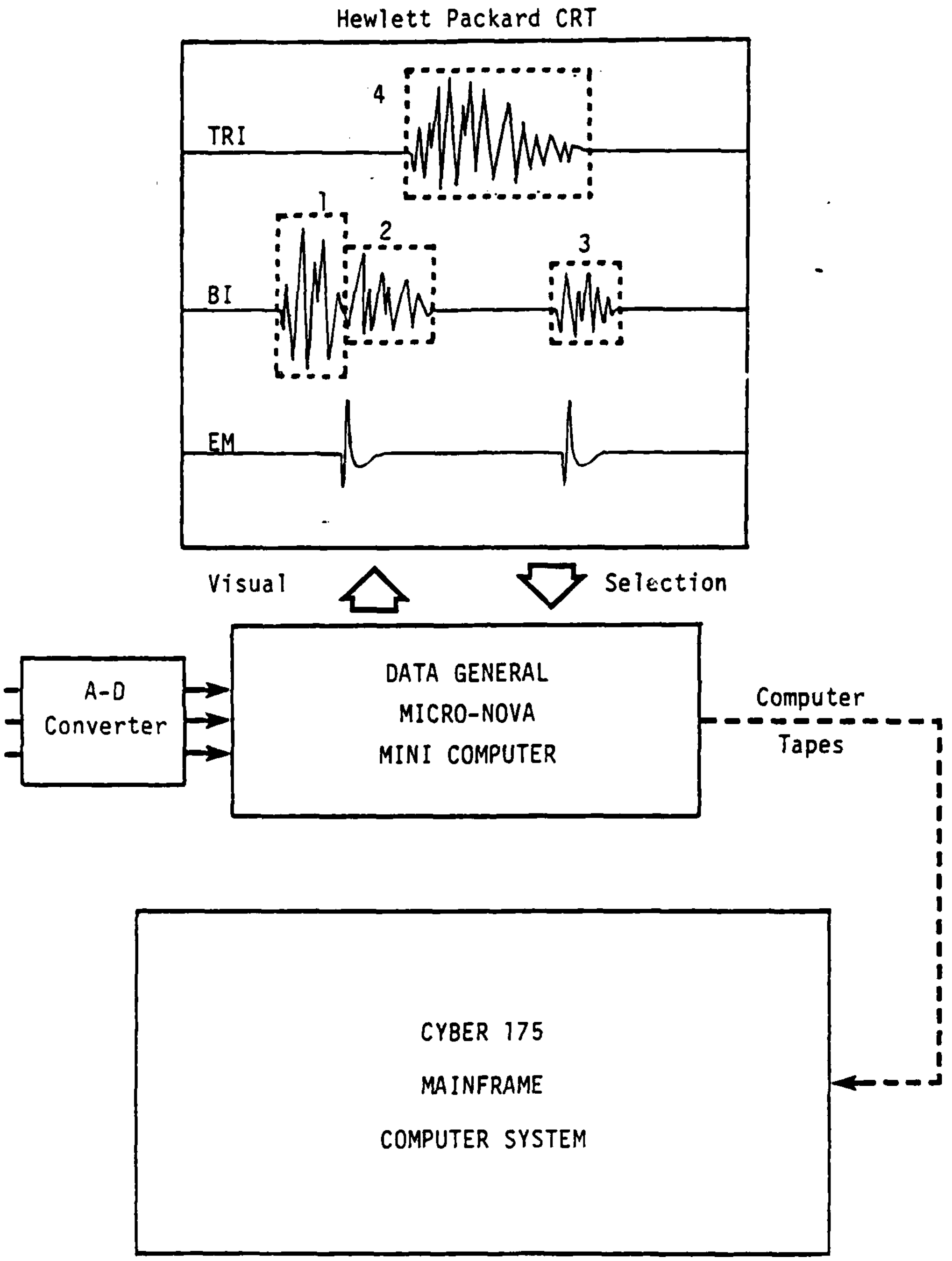


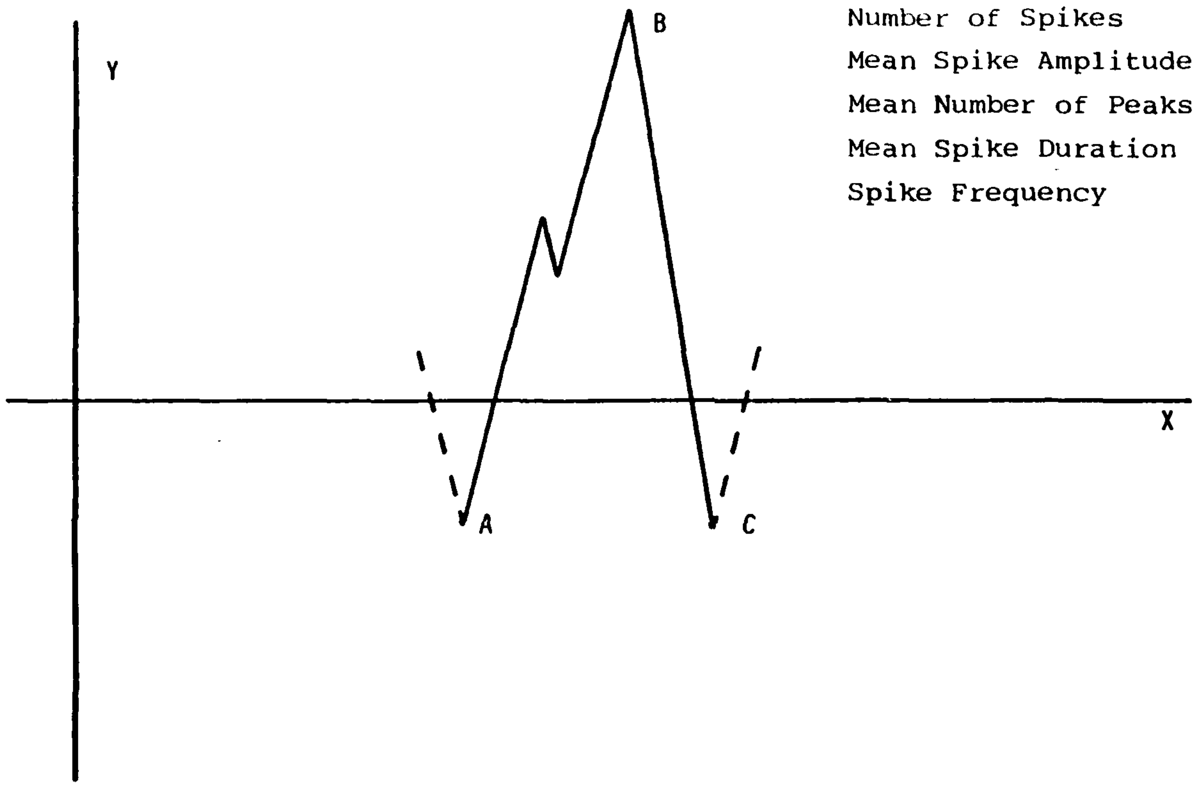
Figure 7. Digital phase of EMG signal quantification 1) biceps brachii motor time, 2) end of the first biceps brachii burst, 3) second biceps brachii burst, and 4) triceps brachii burst digital raw EMG signals

biceps brachii and triceps brachii digital raw EMG signals and the movement start and movement end event markers were simultaneously plotted in proper temporal relation to each other on the Hewlett Packard CRT screen as shown in Figure 7. With the aid of the plotted event markers, each biceps brachii digital raw EMG signal was partitioned into a biceps brachii motor time digital raw EMG signal and an end of the first biceps brachii burst digital raw EMG signal. In addition, a second biceps brachii burst digital raw EMG signal and a triceps brachii digital raw EMG signal were visually selected from the plotted EMG signals. These partitioned biceps brachii digital raw EMG signals as well as the triceps brachii digital raw EMG signal were coded as to subject, day, load, and trial and stored on computer tapes and eventually loaded on to hard disk within a Cyber 175 Computer system to await further reduction. An APL computer program (written by Jean P. Boucher, Dept. of Exercise Science, University of Massachusetts, Amherst) in conjunction with the Cyber 175 Computer System reduced each biceps brachii motor time digital raw EMG signal, end of the first biceps brachii burst digital raw EMG signal, second biceps brachii burst digital raw EMG signal, and triceps brachii burst digital raw EMG signal by calculating the number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike, mean spike slope, and the spike frequency as defined below.

The EMG quantification method that was utilized in this

study incorporates the following basic assumptions. First, each EMG signal is assumed to be a composite of discrete spikes and peaks each shaped by an upward and downward deflection [13,25,40,91,92]. Spikes are differentiated from peaks in that both deflections of a spike cross zero isoelectic baseline [54] and are at least 100uv in amplitude [91]. In contrast, a peak's deflections may or may not cross zero isoelectric baseline or be 100uv in amplitude or greater. However, a peak's deflections are always observed as being part of one or both of the deflections of a spike. In other words, a peak is any pair of upward and downward deflections within a spike that do not together constitute a discrete spike, except in the special case of a spike with a single peak. In this case, the deflections of the spike and the peak are the same . Furthermore, any deflections that do not constitute discrete spikes as previously defined and are found before, between, or after identified spikes are assumed to be background noise and not EMG signal [40,41] .

The following algorithms were used in the calculation of spike parameters and should be related back to Figure 8 for clarity.



Number of Spikes
Mean Spike Amplitude
Mean Number of Peaks Per Spike
Mean Spike Duration
Spike Frequency

Figure 8. Quantification of simplified raw EMG spike. Upward and downward deflections of a spike are greater than 100 uv and cross zero isoelectric baseline. (See parameter algorithms in text for further detail)

Mean Spike Amplitude (MSA) Algorithm

$$(1)\ SA_i = [(B-A)+(B-C)]/2$$

$$(2)\ MSA = \sum_{i=1}^{NS} SA_i/NS$$

where,

SA_i=spike amplitude

A=the first Y coordinate of the upward deflection of a spike within a given sample of raw EMG signal.

B=the last Y coordinate of the upward deflection of a spike within a given sample of raw EMG signal.

C=the last Y coordinate of the downward deflection of a spike within a given sample of raw EMG signal.

NS=the number of spikes within a given sample of raw EMG signal.

Mean Spike Duration (MSD) Algorithm

$$(3)\ SD_i = A - C$$

$$(4)\ MSD = \sum_{i=1}^{NS} SD_i / NS$$

where,

SD_i=spike duration

A=the first X coordinate of the upward deflection of a spike within a given sample of raw EMG signal.

C=the last X coordinate of the downward deflection of a spike within a given sample of raw EMG signal.

NS=the number of spikes within a given sample of raw EMG signal.

Mean Number of Peaks Per Spike (MNPPS) Algorithm

$$(5)\ MNPPS = P_i / NS$$

where,

P_i=the number of sequential pairs of upward and downward deflections between the first X and Y coordinates of the upward deflection and the last X and Y coordinates of the downward deflection of a given spike within a given sample of raw EMG signal.

NS=the number of spikes within a given sample of raw EMG signal.

Mean Spike Slope (MSS) Algorithm

$$(6)\ SS_i=(B-A)/(B_1-A_1)$$

$$(7)\ MSS=\sum_{i=1}^{NS} SS_i/NS$$

where,

SS_i=spike slope

A=the last Y coordinate of the upward deflection of a given spike within a given sample of raw EMG signal.

B=the first Y coordinate of the upward deflection of a given spike within a given sample of raw EMG signal.

A_1=the last X coordinate of the upward deflection of a given spike within a given sample of raw EMG signal.

B_1=the first X coordinate of the upward deflection of a given spike within a given sample of raw EMG signal.

NS=the number of spikes within a given sample of raw EMG signal.

Spike Frequency (SF) Algorithm

$$(8)\ SF=NS/TD$$

where,

NS=the number of spikes within a given sample of EMG signal.

TD=total duration of a given sample of EMG signal.

Design and Statistical Analysis of Data

The data were analyzed with the primary intention of identifying significant differences between test day treatment conditions for each parameter. Identification of reliable EMG spike parameters was important since the inferential statistical procedures utilized in this study assume that parameters are reliable and risk invalidity if unreliable parameters are used. Secondary intentions of this study included assessing sex, load, and trial main treatment effects and the interactions of day, sex, load, and trial main treatment effects.

The parameters that were studied included two kinematic parameters and six EMG spike parameters. The kinematic parameters were maximum speed forearm flexion movement time and maximum speed forearm flexion percent acceleration time. The six EMG spike parameters included number of spikes, mean spike amplitude, mean spike duration, mean spike slope, mean number of peaks per spike, and spike frequency of each biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signal. Data were collected using eight female and eight male subjects while they performed 10 maximal speed forearm flexion trials under three different loads on the first, fourth, and eighth of eight practice days. No data were collected on all other

practice days.

Reliability estimations of the stability of kinematic parameters and EMG spike parameters over test days one, four, and eight were determined by Winer's [94] four-way factorial analysis of variance design with repeated measures. Reliability estimates of the consistency of kinematic parameters and EMG spike parameters over and within test days one, four, and eight were determined using intraclass reliability coefficients [51].

Assessment of main treatment effects and the interaction of main treatment effects for each parameter described above was performed by Winer's [94] four-way factorial analysis of variance design with repeated measures. This design involves the calculation of F-ratios for 1) a sex main effect, 2) a days main effect, 3) a days by sex interaction, 4) a loads main effect, 5) a loads by sex interaction, 6) a days by loads interaction, 7) a days by loads by sex interaction, 8) a trials main effect, 9) a trials by sex interaction, 10) a sex by trials interaction, 11) a days by trials by sex interaction, 12) a loads by trials interaction, 13) a loads by trials by sex interaction, 14) a days by loads by trials interaction, 15) a days by loads by trials by sex interaction. Individual main factor cell mean differences were assessed using Lindquist's [51] Critical Difference Test. In order to assess the relationship between main factors, intercorrelations between all kinematic and EMG spike parameters were determined by Pearson product-moment correlation coefficients [22].

RESULTS

Introduction

The data to be analyzed consisted of maximum speed forearm flexion movement time, maximum speed forearm flexion percent acceleration time, and number of spikes, mean spike amplitude, mean spike duration, mean spike slope, mean number of peaks per spike and spike frequency of each biceps brachii motor time digital raw EMG signal, end of the first biceps brachii burst digital raw EMG signal, second biceps brachii burst digital raw EMG signal, and triceps brachii burst digital raw EMG signal. Sixteen subjects, eight male and eight female, performed 10 maximal speed forearm flexion trials under three different loads on each of eight practice days with at least 30 seconds rest between trials. All parameters were calculated from data collected on test days one, four and eight.

The data were analyzed with the intent of comparing each parameter across test days. Comparisons of each parameter between sexes, loads and trials (within days) were secondary intentions. To meet these intentions, the results are

divided into four main sections: 1) Reliability of kinematic parameters (maximum speed forearm flexion movement time and percent acceleration time) and EMG spike parameters (number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike , mean spike slope, and spike frequency) for each biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signal; 2) Analysis of variance of kinematic parameters; 3) Analysis of variance of EMG spike parameters for each biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii burst digital raw EMG signal; 4) Intercorrelations between EMG spike parameters for each biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signal as well as maximum speed forearm flexion movement time and maximum speed forearm flexion percent acceleration time.

Analysis of the data was conducted according to the procedures outlined **previously.** Reliability of kinematic and EMG spike parameters are presented in terms of their stability and consistency of measurement. Stability over test days was tested by F ratios representing the day main effect obtained using Winer's [94] four-way factorial design analysis of variance with repeated measures on the last three factors. To determine stabilization of all parameters in further detail, Lindquist's

Critical Difference Test [51] was used to assess differences between test day means. The consistency of kinematic and EMG spike parameters was determined using intraclass correlation coefficients. The analysis of test day main effects as well as sex and load main effects and interaction terms for kinematic and EMG spike parameters was determined with the same Winer [94] four-way factorial design that was used to determine stability. Differences between trial means, sex means, and load means for each parameter were assessed with the same critical difference procedure [51] used to assess stabilization of parameters across days. Intercorrelations between EMG spike parameters for each biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signal and maximum speed forearm flexion movement time and maximum speed forearm flexion percent acceleration time were determined by Pearson r correlation coefficients [22]. Prior to the analysis of data, an alpha level of .05 was selected as the criterion for statistical significance. All statistical interpretations were based on this maximum error probability level.

Reliability of Kinematic and EMG Spike Parameters

The data to be analyzed consisted of two kinematic parameters (maximum speed of forearm flexion movement time and percent acceleration time) and six EMG spike parameters (number of spikes, mean spike amplitude, mean spike duration, mean spike slope, mean number of peaks per spike, and spike frequency) measured for each biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii burst digital raw EMG signal. Data were collected on test days one, four, and eight while all other days served only as practice days. On each of the eight practice days, each subject performed ten maximal speed forearm flexions under three different loads. Load zero was no load, while load one was four times the moment of inertia of the forearm and load two was eight times the moment of inertia of the forearm.

Stability and consistency of all parameters across trials within days and across days are important considerations in the present study since treatment effects were to be judged across days. Inherent in the design of this experiment is the paradox that the potency of the days treatment effect may simultaneously weaken the reliability of a particular parameter. The inconsistency of parameters, as indicated by intraclass correlation coefficients, would also confound any pre-post

treatment effects. The reliability analysis was of special significance to the present study since this was a first attempt to measure EMG spike parameters during ballistic muscular contractions in which improvements in execution quality have been assessed.

In the following section, the means and standard deviations of all parameters will be presented followed by the stability of all parameters followed by the consistency of all parameters. Means and standard deviations of all parameters for combined males and females over test days within each load are shown in Table 1. F ratios and mean separation results for day main effects used to determine the stability of all parameters are presented in Table 2. Intraclass correlation coefficients for all parameters are presented in Table 3.

Means and standard deviations of kinematic parameters. Decreases in maximum speed forearm flexion movement time were recorded over test days for combined sex and trial data as shown in Table 1. As shown in Table 1, standard deviations of maximum speed forearm flexion movement time measures decreased over test days one, four, and eight. The greatest decrease in these standard deviations was observed to occur between test days one and four suggesting that subjects not only got faster but became more homogeneous over the first four practice days.

Increases in maximum speed forearm flexion percent

TABLE 1

Means and standard deviations of *maximum* speed forearm flexion movement time and percent acceleration time and number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike, mean spike slope, and spike frequency for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signals across all testing days, loads, and sexes, including combined sexes.

	Load 0			Load 1			Load 2		
Movement Times (ms)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	168.8	156.7	148.9	171.9	173.9	162.2	206.8	205.3	197.3
S.D.	77.8	25.4	21.4	28.7	28.3	19.7	29.1	27.3	23.7
Males (*n*=8)									
Mean	134.6	118.7	114.2	139.9	128.6	123.4	160.6	157.9	149.0
S.D.	30.1	8.3	7.1	38.1	7.4	7.1	13.1	12.3	6.9
Combined (n=16)									
Mean	151.7	137.7	131.6	155.9	151.3	142.8	183.7	181.6	173.2
S.D.	61.4	26.7	23.6	37.2	30.7	24.4	32.3	31.8	29.8
Percent Acceleration Time (%)									
Females (n=8)									
Mean	75.1	81.3	82.6	77.3	80.1	84.3	67.3	71.8	75.7
S.D.	13.0	10.4	12.5	11.7	9.6	9.6	13.6	12.9	11.4
Males (n=8)									
Mean	86.4	89.4	89.8	86.9	90.3	89.2	84.5	86.1	86.9
S.D.	12.0	5.9	8.8	12.3	6.9	9.0	11.3	10.9	9.9
Combined (n=16)									
Mean	80.8	85.4	86.2	82.1	85.2	86.8	75.9	79.0	81.3
S.D.	13.7	9.3	11.4	12.9	9.8	9.6	15.2	13.9	12.0

TABLE 1 (Cont)

Number of Spikes - Biceps Motor Time (spikes)	Load 0			Load 1			Load 2		
	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	8.1	7.1	8.0	8.4	7.1	8.7	8.5	7.9	9.3
S.D.	1.6	1.8	2.2	2.0	1.9	2.0	1.9	2.0	1.9
Males (n=8)									
Mean	6.3	7.2	6.7	6.9	7.5	7.4	8.2	8.5	8.9
S.D.	2.3	2.1	2.1	2.7	2.6	2.0	2.7	2.5	2.7
Combined (n=16)									
Mean	7.2	7.2	7.4	7.6	7.3	8.0	8.3	8.2	8.1
S.D.	2.2	1.9	2.2	2.5	2.3	2.1	2.3	2.3	2.3
Number of Spikes - End Biceps First Burst (spikes)									
Females (n=8)									
Mean	10.0	6.5	8.0	9.0	8.1	8.2	10.3	9.1	10.0
S.D.	5.6	2.1	3.8	4.4	3.6	3.6	4.2	3.2	3.3
Males (n=8)									
Mean	5.4	5.1	5.2	6.0	4.7	5.4	6.5	6.1	6.4
S.D.	2.9	2.0	2.1	3.7	1.6	2.4	2.2	1.5	2.2
Combined (n=16)									
Mean	7.7	5.9	6.6	7.5	6.4	6.8	8.4	7.6	8.2
S.D.	5.0	2.2	3.4	4.3	3.3	3.4	3.8	2.9	3.3

TABLE 1 (Cont)

Number of Spikes - Second	Load 0			Load 1			Load 2		
Biceps Burst (spikes)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	11.8	6.7	7.6	10.1	6.5	7.9	8.4	6.0	8.3
S.D.	4.1	3.4	3.8	3.7	4.0	3.3	3.2	3.4	3.9
Males (n=8)									
Mean	9.0	7.9	7.6	10.1	7.2	8.2	8.1	7.3	7.6
S.D.	5.2	3.4	3.7	6.1	3.5	3.7	4.6	3.7	3.5
Combined (n=16)									
Mean	10.4	6.9	7.6	10.1	6.8	8.0	8.2	6.6	7.9
S.D.	4.9	3.4	3.7	5.0	3.8	3.5	3.4	3.6	3.7
Number of Spikes - Triceps Burst (spikes)									
Females (n=8)									
Mean	22.1	10.7	12.3	23.4	12.6	13.3	19.2	13.0	13.8
S.D.	8.6	4.4	6.5	9.1	5.4	6.2	7.4	5.4	5.2
Males (n=8)									
Mean	8.8	9.0	8.3	9.5	9.2	8.6	10.6	10.8	10.4
S.D.	4.5	3.1	4.0	4.8	2.6	4.4	3.9	3.4	4.2
Combined (n=16)									
Mean	15.5	9.9	10.3	16.4	10.9	10.9	14.9	11.9	12.1
S.D.	9.6	3.9	5.8	10.1	4.6	5.9	7.3	4.6	5.0

TABLE 1 (Cont)

Mean Spike Amplitude -	Load 0			Load 1			Load 2		
Biceps Motor Time (mv)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	.853	.938	1.398	.985	1.003	.977	1.048	1.046	1.076
S.D.	.498	.612	1.687	.822	.753	.784	.813	.861	.788
Males (n=8)									
Mean	.995	1.085	1.212	1.759	1.142	1.208	1.511	1.059	1.297
S.D.	1.275	.500	.926	2.238	.600	.784	2.246	.500	.831
Combined (n=16)									
Mean	.924	1.012	1.305	1.372	1.073	1.093	1.280	1.053	1.187
S.D.	.967	.562	1.360	1.724	.709	.790	1.699	.702	.814
Mean Spike Amplitude - End Biceps First Burst (mv)									
Females (n=8)									
Mean	.902	1.047	1.611	.943	1.015	1.007	1.123	1.218	1.060
S.D.	.669	.709	2.099	.645	.783	.743	.989	1.080	.841
Males (n=8)									
Mean	.922	1.022	1.114	1.289	1.000	.984	1.244	.990	1.140
S.D.	.936	.386	.732	1.482	.592	.507	1.808	.557	.632
Combined (n=16)									
Mean	.912	1.035	1.362	1.116	1.008	.995	1.184	1.104	1.100
S.D.	.811	.569	1.587	1.153	.692	.634	1.454	.864	.742

TABLE 1 (Cont)

Mean Spike Amplitude - Second Biceps Burst (mv)	Load 0			Load 1			Load 2		
	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	.463	.371	.514	.396	.348	.350	.418	.355	.396
S.D.	.226	.229	.396	.219	.169	.138	.330	.179	.193
Males (n=8)									
Mean	.385	.390	.521	.541	.425	.479	.520	.396	.494
S.D.	.252	.157	.247	.564	.232	.199	.563	.182	.213
Combined (n=16)									
Mean	.424	.381	.517	.469	.387	.415	.469	.375	.445
S.D.	.242	.196	.329	.433	.206	.183	.463	.181	.208

Mean Spike Amplitude - Triceps Burst (mv)									
Females (n=8)									
Mean	.751	1.103	1.093	.964	1.166	.793	1.179	1.439	.841
S.D.	.376	.919	.783	.512	.995	.477	.745	1.543	.522
Males (n=8)									
Mean	.659	1.034	.953	.800	.772	.781	.795	.834	.686
S.D.	.395	.757	.575	.695	.382	.455	.735	.333	.296
Combined (n=16)									
Mean	.673	1.068	1.023	.882	.969	.787	.987	1.136	.764
S.D.	.392	.840	.688	.614	.777	.464	.762	1.153	.430

TABLE 1 (Cont)

Mean Spike Duration -	Load 0			Load 1			Load 2		
Biceps Motor Time (ms)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	12.0	11.3	10.6	12.2	11.7	11.2	12.6	11.7	10.8
S.D.	2.5	2.5	2.2	2.7	2.3	2.5	2.4	2.2	2.1
Males (n=8)									
Mean	12.4	10.2	10.9	13.1	11.3	11.0	12.5	11.1	10.7
S.D.	3.5	2.0	2.6	5.2	2.7	2.4	3.9	2.3	2.5
Combined (n=16)									
Mean	12.2	10.8	10.7	12.6	11.5	11.1	12.5	11.4	10.8
S.D.	3.0	2.3	2.4	4.1	2.4	2.4	3.2	2.9	2.3
Mean Spike Duration -									
End Biceps First Burst (ms)									
Females (n=8)									
Mean	13.9	14.6	13.4	14.5	14.5	13.7	14.4	13.2	12.8
S.D.	3.3	3.5	2.9	3.4	3.5	3.1	2.9	2.8	2.1
Males (n=8)									
Mean	16.4	13.7	13.2	15.8	14.2	13.1	14.4	13.1	13.2
S.D.	4.9	3.6	3.4	6.0	3.8	3.5	3.6	2.5	2.6
Combined (n=16)									
Mean	15.2	14.2	13.3	15.2	14.4	13.4	14.4	13.1	13.0
S.D.	4.3	3.6	3.1	4.9	3.6	3.3	3.3	2.7	3.4

TABLE 1 (Cont)

Mean Spike Duration -	Load 0			Load 1			Load 2		
Second Biceps Burst (ms)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	14.1	15.0	13.5	13.8	15.8	14.6	15.3	15.7	13.4
S.D.	3.5	4.0	3.3	1.9	8.8	6.4	3.4	5.7	2.0
Males (n=8)									
Mean	16.0	12.9	12.7	16.0	12.9	12.6	15.2	14.2	12.6
S.D.	5.2	2.5	2.9	5.5	2.5	2.4	4.4	5.2	2.4
Combined (n=16)									
Mean	15.1	14.0	13.1	14.9	14.3	13.6	15.3	14.9	13.0
S.D.	4.5	3.5	3.1	4.2	6.6	4.9	3.9	5.5	2.2
Mean Spike Duration -									
Triceps Burst (ms)									
Females (n=8)									
Mean	12.2	12.1	11.4	11.2	11.5	11.4	11.5	11.6	11.4
S.D.	2.5	2.7	1.9	1.7	2.8	1.8	1.9	2.4	1.8
Males (n=8)									
Mean	14.9	10.2	11.6	13.5	11.2	12.0	13.2	11.6	11.4
S.D.	8.0	1.9	2.1	6.7	2.2	2.5	2.4	2.4	1.8
Combined (n=16)									
Mean	13.5	11.1	11.5	12.3	11.3	11.7	12.4	11.3	11.6
S.D.	6.0	2.5	2.0	5.0	2.5	2.2	2.3	2.1	1.8

TABLE 1 (Cont)

Mean Number of Peaks Per Spike -	Load 0			Load 1			Load 2		
Biceps Motor Time (peaks)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	2.0	2.0	2.0	2.0	2.0	2.1	2.1	2.0	2.0
S.D.	.4	.4	.5	.5	.4	.5	.5	.4	.3
Males (n=8)									
Mean	2.1	1.8	2.0	2.2	1.9	2.0	2.0	2.0	2.0
S.D.	.7	.4	.6	1.0	.4	.5	.5	.5	.5
Combined (n=16)									
Mean	2.0	1.9	2.0	2.1	1.9	2.0	2.1	2.0	2.0
S.D.	.6	.4	.5	.8	.4	.5	.5	.4	.4
Mean Number of Peaks Per Spike -									
End Biceps First Burst (peaks)									
Females (n=8)									
Mean	2.0	2.1	2.0	2.1	2.1	2.1	2.1	2.0	2.1
S.D.	.4	.6	.5	.6	.5	.5	.4	.5	.4
Males (n=8)									
Mean	2.1	1.9	1.9	2.1	2.0	1.8	2.0	1.9	1.9
S.D.	.7	.5	.7	.6	.7	.5	.5	.5	.5
Combined (n=16)									
Mean	2.1	2.0	1.9	2.1	2.1	2.0	2.0	2.0	2.0
S.D.	.6	.5	.6	.6	.6	.5	.5	.5	.4

TABLE 1 (Cont)

Mean Number of Peaks Per Spike - Second Biceps Burst (peaks)	Load 0			Load 1			Load 2		
	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	2.2	2.3	2.0	2.1	2.2	2.2	2.2	2.3	2.0
S.D.	.8	.9	.6	.4	1.3	.8	.6	1.1	.5
Males (n=8)									
Mean	2.3	1.8	1.8	2.4	1.8	1.8	2.2	2.0	1.8
S.D.	1.0	.4	.4	.9	.5	.4	.8	.6	.4
Combined (n=16)									
Mean	2.3	2.0	1.9	2.2	2.0	2.0	2.2	2.1	1.9
S.D.	.9	.7	.5	.7	1.0	.6	.7	.9	.5

Mean Number of Peaks Per Spike - Triceps Burst (peaks)									
Females (n=8)									
Mean	2.1	2.1	2.0	2.1	2.0	2.0	-.1	2.1	2.1
S.D.	.3	.4	.3	.3	.4	.3	.3	.4	.4
Males (n=8)									
Mean	2.3	1.8	1.9	2.1	1.9	1.9	2.1	1.9	2.0
S.D.	1.2	.4	.4	1.2	.4	.4	.5	.4	.3
Combined (n=16)									
Mean	2.2	2.0	1.9	2.1	1.9	1.9	2.1	2.0	2.0
S.D.	.9	.4	.4	.9	.4	.4	.4	.4	.4

TABLE 1 (Cont)

Mean Spike Slope -	Load 0			Load 1			Load 2		
Biceps Motor Time	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	.150	.183	.271	.166	.179	.179	.179	.179	.204
S.D.	.069	.110	.303	.114	.115	.113	.128	.117	.125
Males (n=8)									
Mean	.179	.217	.235	.364	.208	.234	.262	.203	.257
S.D.	.189	.082	.169	.514	.091	.146	.332	.085	.158
Combined (n=16)									
Mean	.165	.200	.253	.265	.194	.207	.220	.191	.231
S.D.	.142	.098	.245	.384	.104	.133	.255	.103	.145
Mean Spike Slope -									
End Biceps First Burst									
Females (n=8)									
Mean	.150	.168	.292	.154	.166	.163	.178	.207	.189
S.D.	.071	.094	.402	.081	.103	.084	.121	.139	.129
Males (n=8)									
Mean	.151	.184	.216	.228	.175	.198	.222	.184	.224
S.D.	.135	.054	.152	.273	.089	.106	.290	.093	.134
Combined (n=16)									
Mean	.150	.176	.254	.191	.170	.181	.200	.195	.207
S.D.	.108	.077	.306	.204	.096	.097	.223	.118	.132

TABLE 1 (Cont)

Mean Spike Slope -	Load 0			Load 1			Load 2		
Second Biceps Burst	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	.082	.061	.096	.070	.057	.061	.067	.059	.070
S.D.	.038	.038	.089	.031	.029	.027	.038	.033	.037
Males (n=8)									
Mean	.071	.077	.102	.102	.080	.092	.092	.075	.091
S.D.	.058	.041	.052	.014	.044	.041	.105	.039	.038
Combined (n=16)									
Mean	.077	.069	.099	.086	.068	.077	.079	.067	.081
S.D.	.049	.040	.073	.085	.039	.038	.079	.037	.038

Mean Spike Slope - Triceps Burst									
Females (n=8)									
Mean	.149	.231	.220	.193	.253	.151	.220	.301	.170
S.D.	.077	.192	.162	.090	.219	.079	.120	.293	.102
Males (n=8)									
Mean	.109	.239	.199	.152	.175	.161	.156	.187	.144
S.D.	.076	.165	.127	.129	.078	.095	.156	.091	.071
Combined (n=16)									
Mean	.129	.235	.209	.172	.213	.156	.188	.244	.157
S.D.	.079	.178	.146	.133	.168	.087	.142	.224	.089

TABLE 1 (Cont)

Spike Frequency - Biceps Motor Time (Hz)	Load 0			Load 1			Load 2		
	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	87.3	92.1	98.2	86.5	88.9	94.3	82.6	88.4	96.0
S.D.	18.6	18.8	20.1	20.8	17.9	21.3	16.0	16.3	19.3
Males (n=8)									
Mean	86.7	101.8	97.1	85.0	94.3	95.0	87.1	94.4	98.6
S.D.	22.8	21.4	22.3	17.8	21.3	19.3	24.8	21.0	22.2
Combined (n=16)									
Mean	87.0	96.9	97.6	85.7	91.3	94.4	84.8	91.4	97.3
S.D.	20.7	20.7	21.2	23.2	20.6	20.3	20.9	18.9	20.8
Spike Frequency - End Biceps First Burst (Hz)									
Females (n=8)									
Mean	75.6	72.0	78.1	72.6	72.9	76.5	72.2	78.9	80.5
S.D.	16.4	15.6	16.0	17.3	17.4	15.6	14.1	15.8	14.0
Males (n=8)									
Mean	65.9	78.1	80.2	69.4	74.8	81.9	73.7	79.5	77.9
S.D.	18.0	21.2	19.1	20.0	17.6	21.4	17.5	15.3	16.4
Combined (n=16)									
Mean	70.7	75.0	79.2	71.0	73.8	79.2	73.0	79.2	79.2
S.D.	17.8	18.8	17.5	18.6	17.5	18.9	15.8	15.5	15.3

TABLE 1 (Cont)

Spike Frequency -	Load 0			Load 1			Load 2		
Second Biceps Burst (Hz)	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8	Day 1	Day 4	Day 8
Females (n=8)									
Mean	74.9	70.0	77.9	74.0	69.5	74.6	68.4	69.3	76.5
S.D.	16.3	14.9	16.0	10.5	14.8	16.4	13.5	16.4	12.0
Males (n=8)									
Mean	68.0	80.3	82.4	70.8	80.5	82.2	70.4	77.3	81.9
S.D.	17.1	14.8	17.8	19.1	13.9	16.7	15.5	19.5	14.8
Combined (n=16)									
Mean	71.4	75.2	80.1	72.4	75.0	78.4	69.4	73.3	79.2
S.D.	17.0	15.6	17.0	15.4	15.3	16.9	14.5	18.4	13.7
Spike Frequency -									
Triceps Burst (Hz)									
Females (n=8)									
Mean	84.3	86.9	89.8	91.6	90.7	90.3	89.5	90.5	90.1
S.D.	15.3	19.2	14.6	14.5	20.8	14.8	16.8	20.9	14.0
Males (n=8)									
Mean	75.1	101.8	89.0	80.5	92.8	86.7	78.2	92.9	86.4
S.D.	18.6	18.5	16.1	18.5	18.0	16.3	13.4	14.4	12.9
Combined (n=16)									
Mean	79.7	94.3	89.4	86.0	91.8	88.5	83.8	91.7	88.2
S.D.	17.6	20.2	15.4	17.5	19.4	15.6	16.2	17.2	13.5

acceleration time were observed for combined sex and trial data over test days as shown in Table 1. Standard deviations of maximum speed forearm flexion percent acceleration time ranged from 9.3 to 15.2 and decreased between test days one and four as shown in Table 1.

Means and standard deviations of EMG spike parameters. Decreases were observed in number of spikes means between test day one and test day four for combined sex and trial data, shown in Table 1, for end of the first biceps brachii burst , second biceps brachii burst , and triceps brachii burst digital raw EMG signals. Table 1 shows that test day one standard deviations for number of spikes are generally greater than test day four standard deviations for number of spikes for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii burst digital raw EMG signal. Standard deviations for number of spikes generally stayed the same or increased slightly between test days four and eight but test day eight standard deviations rarely exceeded test day one standard deviations (see Table 1).

Mean spike amplitude means for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii burst digital raw EMG signals ranged from as low as .375mv to as high as 1.372 mv as shown for combined groups in Table 1. Mean spike amplitude was generally

smaller for the second biceps brachii burst digital raw EMG signal than for the three other digital raw EMG signals as shown in Table 1. Standard deviations were large for mean spike amplitude measures, being often as large or larger than mean spike amplitude means. As can be seen in Table 1, mean spike amplitude standard deviations decreased between test days one and four for all digital raw EMG signals except triceps brachii burst digital raw EMG signals.

Mean spike duration for biceps brachii motor time digital raw EMG signals for combined sexes and trials, shown in Table 1, was observed to decrease between test day one and test day four. Similar decreases were observed for end of the first biceps brachii burst digital raw EMG signals between test day one and test day four, for second biceps brachii burst digital raw EMG signal between test day one and test day four, and for triceps brachii burst digital raw EMG signal between test day one and test day four. As shown in Table 1, mean spike duration standard deviations decreased between test days one and four for all digital raw EMG signals except second biceps brachii burst digital raw EMG signals under loads two and three.

Means and standard deviations for mean number of peaks per spike are presented in Table 1. It is obvious that mean number of peaks per spike means for combined sexes resulted in a narrow range of 1.9 to 2.3 peaks per spike over test days one, four, and eight over all loads. Mean number of peaks per spike standard

deviations either decreased or remained constant between test days one and four for all digital raw EMG signals except second biceps brachii burst digital raw EMG signal as shown in Table 1.

Mean spike slope was generally lower for second biceps brachii burst digital raw EMG signal than for biceps brachii motor time, end of the first biceps brachii burst and triceps brachii burst digital raw EMG signals as can be seen in Table 1. Mean spike slope standard deviations were high, occasionally exceeding mean values as shown in Table 1.

Spike frequency means were, without exception, greater on test day four than test day one for all load conditions for all biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signals as can be seen in Table 1. Spike frequency for biceps brachii motor time (84.8 Hz to 97.3 Hz) and triceps brachii burst (79.7 Hz to 94.3 Hz) digital raw EMG signals were similar and both generally greater than the similar spike frequency for second biceps brachii burst (69.4 Hz to 80.1 Hz) and end of the first biceps brachii burst (70.7 Hz to 79.2 Hz) digital raw EMG signals. Spike frequency standard deviations were relatively constant across days with biceps brachii motor time digital raw EMG signals showing the greatest standard deviations by being in a 20 to 30 hertz range, with all other digital raw EMG signals falling into the 10 to 20 hertz range.

TABLE 2

Summary table of F values and mean separation results[1] for test day main effects[2] for maximum speed forearm flexion movement time and percent acceleration time and six raw EMG spike parameters for biceps brachii motor time, end of the first biceps brachii burst, second biceps burst, and triceps brachii burst digital raw EMG signals.

Parameter	F ratio	Day Mean Separation[3]		
Movement Time (ms)	7.54**	D8	D4	D1
Percent Acceleration Time (%)	.06			
Number of Spikes				
Biceps Brachii Motor Time	3.04			
End of the First Biceps Brachii Burst	2.78			
Second Biceps Brachii Burst	8.95**	D4	D8	D1
Triceps Brachii Burst	11.61**	D4	D8	D1
Mean Spike Amplitude (mv)				
Biceps Brachii Motor Time	.18			
End of the First Biceps Brachii Burst	.12			
Second Biceps Brachii Burst	.76			
Triceps Brachii Burst	.89			
Mean Spike Duration (ms)				
Biceps Brachii Motor Time	10.35**	D8	D4	D1
End of the First Biceps Brachii Burst	6.53**	D8	D4	D1
Second Biceps Brachii Burst	3.15			
Triceps Brachii Burst	4.75*	D4	D8	D1

Stability of kinematic parameters. The F ratio of 7.54 observed between maximum speed forearm flexion movement time measures over test days shown in Table 2, exceeded the F ratio of 5.45 required for significance at the .01 level with 2 and 28 degrees of freedom. Mean separation procedures showed that test day one maximum speed forearm flexion movement time was significantly longer than test day eight maximum speed forearm flexion movement time, indicating an overall day-practice treatment effect. Experimental days four and eight maximum speed forearm flexion movement time means were not significantly different indicating a stabilization of maximum speed forearm flexion movement time measures over test days four and eight.

The F ratio (F=0.06) for maximum speed forearm flexion percent acceleration time representing the day main treatment effect failed to exceed the F ratio of 3.34 necessary for significance at the .05 level with 2 and 28 degrees of freedom. Thus, it appears that maximum speed forearm flexion percent acceleration time had been stable across all test days.

Stability of EMG spike parameters. As shown in Table 2, F ratios representing the test days main treatment effect for number of spikes for biceps brachii motor time and the end of the first biceps brachii burst digital raw EMG signals failed to reach the value of 3.34 necessary to attain significance ($p \leq .01$). Number of spikes measured for the second biceps brachii burst and

triceps brachii burst digital raw EMG signals resulted in F ratios of 8.95 and 11.61 respectively, both exceeding the minimum value of 5.45 required for significance at the .01 level. Mean separation procedures performed revealed that test day four and test day eight means were not significantly different while both test day four and test day eight means were significantly different from test day one. Thus, the number of spikes parameter for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals appear to have been stable across all days while the number of spikes parameter for second biceps brachii burst and triceps brachii burst digital raw EMG signals stabilized after test day four.

The F ratios for test days main treatment effect for mean spike amplitude for biceps brachii motor time ($F=0.18$), end of the first biceps brachii burst ($F=0.12$), the second biceps brachii burst ($F=0.76$) and the triceps brachii burst ($F=0.89$) digital raw EMG signals failed to attain the F of 3.34 required for significance at the .05 level. In light of this evidence, mean spike amplitude was stable across all test days when measured for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signals.

F ratios for mean spike duration measured for biceps brachii motor time ($F=10.35$), end of the first biceps brachii burst ($F=6.53$) and the triceps brachii burst ($F=4.75$) digital raw EMG

signals all exceeded the F value of 3.34 required for significance at the .05 level. The results of mean separation procedures shown in Table 2 revealed that mean spike duration for the three digital raw EMG signals on test day four were not significantly different from test day eight. Thus, mean spike duration for biceps brachii motor time, end of the first biceps brachii burst, and triceps brachii burst digital raw EMG signals seem to have stabilized after test day four while mean spike duration for the second biceps brachii burst digital raw EMG signal was stable across all test days.

F ratios representing test day main treatment effects for mean number of peaks per spike for biceps brachii motor time ($F=2.07$) and the end of the first biceps brachii burst ($F=1.53$) digital raw EMG signals were not significant at the .05 level. However, test day main treatment effects for mean number of peaks per spike for the second biceps brachii burst ($F=4.57$) and the triceps brachii burst ($F=3.42$) digital raw EMG signals were both significant at the .05 level. Mean separation procedures showed that test day four means and test day eight means for the second biceps brachii burst and triceps brachii burst digital raw EMG signals were not significantly different. Thus, it appears that mean number of peaks per spike measures for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals were stable across all test days while mean number of peaks per spike measures for second biceps brachii burst and

triceps brachii burst digital raw EMG signals stabilized after test day four.

Inspection of the F ratios in Table 2 indicated that the test day main treatment effects for mean spike slope for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii burst digital raw EMG signals were not significant ($p \leq .05$). Therefore, the results indicate that mean spike slope was stable across all test days.

As shown in Table 2, F ratios representing test day treatment effects for spike frequency of biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signals were all significant. Spike frequency test day four means and test day eight means were not significantly different as shown in the mean separation results in Table 2, suggesting that spike frequency for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signals stabilized after test day four.

<u>Consistency of kinematic parameters</u>. In the previous stability analysis of kinematic parameters, the results showed that maximum speed forearm flexion movement time and percent acceleration time were stable over test days four and eight. These findings were expected since Wolcott [95] previously reported similar findings. Since both kinematic parameters were found to be stable after

test day four, it is logical that the following consistency analysis of kinematic parameters incorporate only test days four and eight.

Intraclass correlation coefficients and variance components for kinematic parameters are shown in Table 3. Intraclass reliability coefficients for maximum speed forearm flexion movement time were high (load 0, r=.97; load 1, r=.93; load 2, r=.93). Inspection of variance components revealed that trial error variance (S^2_t=65.07, S^2_t=55.29, S^2_t=96.23) and day error variance (S^2_d=28.28, S^2_d=97.93, S^2_d=107.69) for loads zero, one and two respectfully, were greater under larger loads. Larger true score variance components observed under higher loads (load 0, S^2_t=582.57; load 1, S^2_t=673.94; load 2, S^2_t=810.94) served to offset the potential weakening of the observed intraclass correlation coefficients by higher error variance components at higher loads.

Intraclass correlation coefficients for maximum speed forearm flexion percent acceleration time were acceptable (load 0, r=.70; load 1, r=.64; load 2, r=.86). Inspection of variance components presented in Table 3 showed that more error variance was between trials rather than between days.

Consistency of EMG spike parameters. The previously presented stability analysis of EMG spike parameters showed that all six EMG spike parameters for all four digital raw EMG signals were

TABLE 3

Variance estimates and intraclass reliability coefficients[1] for maximum speed forearm flexion movement time, percent acceleration time, number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike, mean spike slope and spike frequency for biceps brachii motor time (B1), end of the first biceps brachii burst (B2), second biceps brachii burst (B3), and triceps brachii burst (B4) digital raw EMG signals across loads for combined test days and sexes.

	Load 0				Load 1				Load 2			
Measure	S^2Trials	S^2Days	S^2True Score	R	S^2Trials	S^2Days	S^2True Score	R	S^2Trials	S^2Days	S^2True Score	R
Movement Time	65.07	28.28	582.57	.97	55.29	97.93	673.94	.93	96.23	107.69	810.94	.93
% Acceleration Time	44.20	28.40	38.67	.70	36.54	30.00	30.29	.64	60.85	24.28	91.16	.86
Number of Spikes												
B1	2.55	.64	1.20	.73	2.56	.79	1.64	.76	2.79	1.22	1.63	.68
B2	5.01	.84	2.52	.79	6.08	1.00	4 21	.84	5.51	1.41	3.10	.76
B3	7.50	2.20	3.28	.69	7.60	2.84	3.36	.65	7.22	3.56	3.18	.60
B4	14.25	5.49	4.73	.58	15.23	4.70	8.33	.73	14.39	6.41	2.80	.42
Mean Spike Amplitude												
B1	.160	.769	.205	.34	.113	.108	.365	.86	.103	.173	.330	.78
B2	.314	1.042	.124	.19	.122	.121	.212	.76	.141	.144	.390	.83
B3	.032	.044	.003	.12	.020	.007	.012	.72	.018	.012	.010	.59
B4	.194	.274	.136	.48	.110	.213	.106	.49	.193	.465	.154	.39
Mean Spike Duration												
B1	3.87	.62	1.15	.70	3.53	.53	2.17	.83	3.10	.33	2.12	.87
B2	7.83	.80	3.04	.79	8.24	1.46	2.80	.71	4.44	.89	1.24	.67
B3	7.47	2.46	1.15	.42	28.50	3.24	1.96	.39	11.52	5.67	1.59	.32
B4	3.20	1.80	.29	.22	3.72	1.36	.57	.40	2.58	1.18	.09	.11

TABLE 3 (Cont)

Measure	Load 0				Load 1				Load 2			
	S^2Trials	S^2Days	S^2TrueScore	R	S^2Trials	S^2Days	S^2TrueScore	R	S^2Trials	S^2Days	S^2TrueScore	R
Mean Number of Peaks Per Spike												
B1	.196	.028	.023	.48	.178	.006	.019	.61	.134	.000	.049	.86
B2	.289	.000	.026	.76	.248	.060	.001	.03	.191	.003	.042	.79
B3	.340	.010	.049	.66	.587	.038	.070	.57	.380	.050	.070	.61
B4	.131	.020	.017	.50	.131	.000	.017	.73	.119	.012	.010	.47
Mean Spike Slope												
B1	.0033	.0286	.0047	.24	.0027	.0048	.0073	.74	.0025	.0060	.0082	.72
B2	.0095	.0403	.0024	.10	.0029	.0051	.0017	.38	.0012	.0024	.0001	.13
B3	.0012	.0024	.0002	.13	.0007	.0004	.0005	.68	.0006	.0005	.0004	.60
B4	.0068	.0150	.0056	.42	.0031	.0129	.0032	.32	.0064	.0200	.0053	.34
Spike Frequency												
B1	303.79	64.50	74.48	.61	236.23	40.78	151.95	.83	219.12	39.29	155.08	.84
B2	218.01	22.87	99.34	.81	225.21	44.72	72.56	.68	166.26	32.75	40.10	.62
B3	163.27	84.83	28.19	.36	177 29	35.22	54.87	.67	162.40	55.07	58.66	.62
B4	217.97	107.77	5.36	.08	223.92	58.56	33.71	.45	165.74	85.81	5.30	.09

Calculated over test days four and eight.

stable across test days four and eight. Since EMG spike parameters were found to stabilize across test days four and eight, the following consistency analysis incorporated data collected only on test days four and eight. Intraclass reliability coefficients and variance components for all EMG spike parameters under all load conditions are presented in Table 3.

Intraclass reliability coefficients for number of spikes for biceps brachii motor time digital raw EMG signals were acceptable (load 0, r=.73; load 1, r=.76; load 2, r=.68). Intraclass reliability coefficients for number of spikes for end of the first biceps brachii burst digital raw EMG signals (load 0, r=.79; load 1, r=.84; load 2, r=.76) were greater than those for biceps brachii motor time digital raw EMG signals. Lower than either biceps brachii motor time or end of the first biceps brachii burst digital raw EMG signals intraclass correlation coefficients, intraclass correlation coefficients for second biceps brachii burst digital raw EMG signals (load 0, r=.69; load 1, r=.65; load 2, r=.60) were not considered acceptable. Triceps brachii burst digital raw EMG signal intraclass reliability coefficients (load 0, r=.58; load 1, r=.73; load 2, r=.42) were also judged unacceptable for the present study.

Inspection of variance components revealed that number of spikes for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii

burst digital raw EMG signals displayed greater trial variance than day variance implying less consistency between trials than between days. Future attempts at improving reliability of this parameter should therefore focus on increasing the number of trials. This is particularly recommended for number of spikes for the triceps brachii burst digital raw EMG signal since trial variance was more than twice as large for the triceps brachii burst digital raw EMG signal than for any of the other three digital raw EMG signals.

Intraclass reliability coefficients for mean spike amplitude under load zero for biceps brachii motor time ($r=.34$), end of the first biceps brachii burst ($r=.19$), the second biceps brachii burst ($r=.12$), and the triceps brachii burst ($r=.48$) digital raw EMG signals were considerably smaller than under load one ($r=.86$, $r=.76$, $r=.72$, $r=.49$) and load two ($r=.78$, $r=.83$, $r=.53$, $r=.39$) respectively. Based on these intraclass correlation coefficients, mean spike amplitude was judged consistent for only biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals. Examination of variance components shown in Table 3, revealed that day variance under load zero for biceps brachii motor time ($S^2_d=.769$) and end of the first biceps brachii burst ($S^2_d=1.042$) digital raw EMG signals were three to eight times greater than under load one ($S^2_d=.108$, $S^2_d=.121$) and load two ($S^2_d=.173$, $S^2_d=.144$). It is apparent, therefore, that the low intraclass reliability coefficients

observed for mean spike amplitude for biceps brachii motor time ($r=.34$) and end of the first biceps brachii burst ($r=.19$) digital raw EMG signals may have been partially the result of an across test days practice effect. In addition, the true score variance under load zero for biceps brachii motor time ($S^2_t=.205$) and end of the first biceps brachii burst ($S^2_t=.124$) digital raw EMG signals as compared to load one ($S^2_t=.365$, $S^2_t=.212$) and load two ($S^2_t=.330$, $S^2_t=.390$) further explains the observed differences in intraclass reliability coefficients between unloaded (load 0) and loaded (load 1 and 2) conditions.

Intraclass reliability coefficients for mean spike duration for biceps brachii motor time (load 0, $r=.70$; load 1, $r=.83$; load 2, $r=.87$) and end of the first biceps brachii burst (load 0, $r=.79$; load 1, $r=.71$; load 2, $r=.67$) digital raw EMG signals were considerably larger than those for the second biceps brachii burst (load 0, $r=.42$; load 1, $r=.39$; load 2, $r=.32$) and triceps brachii burst (load 0, $r=.22$; load 1, $r=.40$; load 2, $r=.11$) digital raw EMG signals. Therefore, mean spike duration was judged consistent for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals only.

Inspection of the variance components for mean spike duration shown in Table 3 revealed that the major source of error variance was from trial to trial rather than day to day for all four digital raw EMG signals. Also of interest is that while the

low intraclass correlation coefficients for second biceps brachii burst digital raw EMG signals were mostly due to high trial error variance the low intraclass correlation coefficients for triceps brachii burst digital raw EMG signals were predominantely due to large decreases in true score variance.

Intraclass reliability coefficients for mean number of peaks per spike for biceps brachii motor time (load 0, r=.48; load 1, r=.61; load 2, r=.86) and end of the first biceps brachii burst (load 0, r=.76; load 1, r=.03; load 2, r=.79) digital raw EMG signals were acceptable; the second biceps brachii burst (load 0, r=.66; load 1, r=.57; load 2, r=.61) and triceps brachii burst (load 0, r=.50; load 1, r=.73; load 2, r=.47) intraclass reliability coefficients were judged unacceptable. Examination of the variance components shown in Table 3 showed that trial error variance was by far the largest source of error variance for mean number of peaks per spike. Improvements in the reliability of this spike parameter should focus, therefore, on increasing the number of trials.

Intraclass reliability coefficients for mean spike slope under load zero for biceps brachii motor time (r=.24), end of the first biceps brachii burst (r=.10), second biceps brachii burst (r=.13), and triceps brachii burst (r=.42) digital raw EMG signals were all low and judged unacceptable primarily because day error variances were much greater than true score variances (see Table 3). These data suggest that improvements in the

reliability of this spike parameter under unloaded (load 0) conditions may be possible if measurements were made over more days.

Intraclass reliability coefficients for mean spike slope under load one and load two were larger than under load zero for biceps brachii motor time (load 1, $r=.74$; load 2, $r=.72$) and the second biceps brachii burst (load 1, $r=.68$; load 2, $r=.60$). Review of variance components shown in Table 3 revealed that observed improvements in intraclass correlation coefficients under load one and load two were due to lower day error variances as compared to the load zero condition. Intraclass reliability coefficients for mean spike slope for end of the first biceps brachii burst (load 1, $r=.38$; load 2, $r=.13$) and triceps brachii burst (load 1, $r=.32$; load 2, $r=.34$) digital raw EMG signals during load one and load two were more similar to intraclass reliability coefficients for load zero. Based upon the intraclass correlation coefficients presented, the consistency of mean spike slope was judged to be poor.

Intraclass reliability coefficients for spike frequency for biceps brachii motor time digital raw EMG signals (load 0, $r=.61$; load 1, $r=.83$; load 2, $r=.84$) were better for load one and two while the opposite was observed for spike frequency recorded for end of the first biceps brachii burst digital raw EMG signals (load 0, $r=.81$; load 1, $r=.68$; load 2, $r=.62$). These intraclass

reliability coefficients for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals were judged acceptable. Intraclass reliability coefficients for spike frequency for second biceps brachii burst digital raw EMG signals (load 0, r=.36; load 1, r=.67; load 2, r=.62) were lower during load zero than loads one and two and were judged generally unacceptable. Intraclass reliability coefficients for triceps brachii burst digital raw EMG signals (load 0, r=.08; load 1, r=.45; load 2, r=.09) were low for all loads and also were unacceptable.

Inspection of variance components for spike frequency shown in Table 3 revealed that trial error variance was by far the largest source of error variance. This observation suggests that improvements in the intraclass reliability coefficients for all load conditions might be possible if a greater number of trials is planned.

Summary. In this section, the means and standard deviations of all parameters were presented in Table 1. Evidence for the stability of all parameters was obtained from F ratios, presented in Table 2, representing the test day main treatment effect using Winer's [94] four-way factorial analysis of variance with repeated measures. Evidence for the consistency of all parameters was obtained from intraclass reliability coefficients and variance components for each load condition shown in Table 3. Analysis of test days main treatment effects and test day mean

separation procedures revealed that all parameters were stable between test days four and eight. Since all parameters were stable, all parameters that were consistent were deemed reliable.

It is evident that some of the EMG spike parameters were unreliable. This finding is valuable itself since no comparable data exists for EMG spike parameters during gross motor learning. The following parameters were deemed reliable: 1) maximum speed forearm flexion movement time; 2) maximum speed forearm flexion percent acceleration time; 3) number of spikes for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals; 4) mean spike amplitude for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals; 5) mean spike duration for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals; 6) mean number of peaks per spike for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals; 7) spike frequency for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals.

The following parameters were deemed unreliable: 1) number of spikes for second biceps brachii burst and triceps brachii burst digital raw EMG signals; 2) mean spike amplitude for second biceps brachii burst and triceps brachii burst digital raw EMG signals; 3) mean spike duration for second biceps brachii burst and triceps brachii burst digital raw EMG signals; 4) mean spike

slope for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst and triceps brachii burst digital raw EMG signals; 5) spike frequency for second biceps brachii burst and triceps brachii burst digital raw EMG signals.

Analysis of Variance of Kinematic Parameters

Introduction. The measures to be analyzed consisted of maximum speed forearm flexion movement time and percent acceleration time, as well as number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike and spike frequency for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals. Ten trials of maximum speed forearm flexion were performed and all parameters were recorded, under three different load conditions on each of three test days (practice days 1, 4, and 8).

The analysis of variance consisted of Winer's [94] four-way factorial design with repeated measures. This design entails calculation of F ratios for 1) a sex main effect, 2) a days main effect, 3) a days by sex interaction, 4) a loads main effect, 5) a loads by sex interaction, 6) a days by loads interaction, 7) a days by loads by sex interaction, 8) a trials main effect, 9) a trials by sex interaction, 10) a days by trials interaction, 11) a days by trials by sex interaction, 12) a loads by trials interaction, 13) a loads by trials by sex interaction, 14) a days by loads by trials interaction, 15) a days by loads by trials by sex interaction. Individual mean differences were ascertained by Linquist's Critical Difference Test [51]. An alpha level of .05 was chosen as the required level for significance.

Maximum speed forearm flexion movement time. Means and standard deviations of maximum speed forearm flexion movement times for test days 1, 4, and 8 and combined days are presented in Table 4. Included in Table 4 are maximum speed forearm flexion movement time means and standard deviations for males and females, loads zero, one, and two, as well as for combined sexes, combined loads and combined trials. Means and standard deviations for trial one through trial ten for test days one, four, and eight and for combined days, sexes, and loads are shown in Table 5. The four factor analysis of variance of maximum speed forearm flexion movement time for days, sexes, loads, and trials is presented in Table 6.

Main effect: test days 1, 4, 8. Maximum speed forearm flexion movement times, for combined sexes, loads, and trials decreased over test days one (163.8 ms), four (156.9 ms), and eight (149.2 ms). With 2 and 28 degrees of freedom, an F of 5.49 is required for significance at the .01 level. The observed F for between test days of 7.54 exceeded this value necessary for significance. Thus, there were significant differences between maximum speed forearm flexion movement times over the three test days. Mean separation procedures outlined by Lindquist [51] revealed that differences between maximum speed forearm flexion movement times for test days one, four, and eight were

TABLE 4

Means and standard deviations of maximum speed forearm flexion movement times (ms) under three loads and combined loads for females and males and combined sexes over test days one, four, and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	168.8	77.8	171.9	28.7	206.8	29.1	182.5	53.3
Day 4	156.7	25.4	173.9	28.3	205.3	27.3	178.6	33.6
Day 8	148.9	21.4	162.2	19.7	197.3	23.7	169.5	29.7
Combined Days	158.1	49.2	169.3	26.3	203.1	27.0	176.9	40.5
Males								
Day 1	134.6	30.1	139.9	38.1	160.6	13.1	145.0	31.3
Day 4	118.7	8.3	128.6	7.4	157.9	12.3	135.1	19.2
Day 8	114.2	7.1	123.4	7.1	149.0	6.9	128.9	16.3
Combined Days	122.5	20.8	130.7	23.7	155.8	12.1	136.3	24.1
Combined Sexes								
Day 1	151.7	61.4	155.9	37.2	183.7	32.3	163.8	47.5
Day 4	137.7	26.7	151.3	30.7	181.6	31.8	156.9	35.0
Day 8	131.6	23.6	142.8	24.4	173.2	29.8	149.2	31.4
Combined Days	140.3	41.7	150.0	31.6	179.5	31.6		

significant at the .01 level. Decreases in maximum speed forearm flexion movement time due to practice across days is in agreement with previously reported findings [47,95]. Standard deviations of maximum speed forearm flexion movement times for combined sexes, combined loads, and combined trials were greatest on test day one (47.5) and lower on test days four (35.0) and eight (31.4).

Main effect: trials. Maximum speed forearm flexion movement time means for combined test days, sex, and loads shown in Table 5, were greatest for trial one (163.8 ms) and least for trial ten (152.8 ms). With 9 and 126 degrees of freedom, an F of 2.56 is required for significance at the .01 level. The observed F of 4.12 exceeded this value necessary for significance. Thus, significant differences existed between maximum speed forearm flexion movement time means for trials one through ten. The results of mean separation procedures [51] are presented in Table 7. Maximum speed forearm flexion movement times for trial one were significantly different from all other trials. While, considerable overlap occurred among the mean values for trials two through ten, the tendency was for earlier mean values for trials to be significantly different from later mean values for trials. These findings are in agreement with data previously reported by Lagasse [47]. Standard deviations for maximum speed forearm flexion movement times shown in Table 5 were greater for trial one (64.3) than trial two (36.1) through trial ten (34.3).

TABLE 5

Means and standard deviations for maximum speed forearm flexion movement times (ms) for across trials and test days for combined loads and sexes and for combined days, loads, and sexes.

	TRIALS									
Day 1	1	2	3	4	5	6	7	8	9	10
Mean	179.4	169.6	168.4	164.4	166.6	158.5	159.4	158.9	157.6	155.0
s.d.	98.7	40.0	49.5	35.5	33.6	34.5	36.3	38.0	34.7	34.7
Day 4										
Mean	161.1	156.8	157.7	155.0	156.8	157.7	158.5	154.9	154.8	155.3
s.d.	38.2	34.5	33.6	34.6	34.6	35.2	38.1	33.5	34.1	35.6
Day 8										
Mean	150.9	148.8	150.6	150.7	149.3	149.6	146.8	147.0	150.1	148.0
s.d.	30.8	30.8	30.8	35.6	31.2	33.2	29.1	31.1	30.7	32.9
Combined Days										
Mean	163.8	158.4	158.9	156.7	157.6	155.3	154.9	153.6	154.1	152.8
s.d.	64.3	36.1	39.2	35.5	33.7	34.3	35.0	34.4	33.1	34.3

TABLE 6

Analysis of variance for maximum speed forearm flexion movement time

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	35312986.80625	1156.13
SEX	1	591340.80625	19.36**
ERROR	14	30544.16498	
DAYS	2	25602.45625	7.54**
DS	2	1108.76458	.33
ERROR	28	3397.65486	
LOAD	2	199804.55833	318.64**
LS	2	4423.35833	7.05**
ERROR	28	627.05516	
DL	4	1749.02083	2.11*
DLS	4	438.17292	.53
ERROR	56	828.88299	
TRIAL	9	1530.80471	4.12**
TS	9	426.13187	1.15
ERROR	126	371.83054	
DT	18	704.18619	1.75*
DTS	18	410.53927	1.02
ERROR	252	403.15674	
LT	18	446.93179	1.20
LTS	18	529.81358	1.42
ERROR	252	373.34870	
DLT	36	369.77160	.86
DLTS	36	372.72307	.86
ERROR	504	431.94042	

* Significant at the .05 level

** Significant at the .01 level

Main effect:sex. Maximum speed forearm flexion movement times for combined test days, loads, and trials were greater for females (176.9 ms) than for males (136.3 ms). With 1 and 14 degrees of freedom, an F of 8.86 is required for significance at the .01 level. The observed F of 19.36 for between males and females exceeded this value necessary for significance. Therefore, maximum speed forearm flexion movement times were significantly different between sexes. Significantly longer maximum speed forearm flexion movement times for females than males has not been previously reported in the literature. Standard deviations of maximum speed forearm flexion movement times for combined test days, loads and trials were also larger for females (40.5) than males (24.1).

Main effect: loads. Maximal speed forearm flexion movement times for combined test days, sexes, and trials were less for load 0 (140.3 ms) than load 1 (150.0 ms) than load 2 (179.5 ms). With 2 and 28 degrees of freedom an F of 5.49 is required for significance at the .01 level. The observed F for between loads 0, 1, and 2 of 318.64 exceeded the value necessary for significance. Maximum speed forearm flexion movement times were therefore significantly different between loads 0, 1, and 2. Mean separation procedures [51] revealed that maximum speed forearm flexion movement times were significantly different between all three loads at the .01 level. This finding has not previously been reported. Standard deviations of maximum speed

TABLE 7

Lindquist's critical difference test applied to trial mean differences for combined test days, sexes, and loads for maximum speed forearm flexion movement time and mean spike amplitude for biceps motor time digital raw EMG signals.

TRIALS

1 3 2 5 4 6 7 9 8 10

forearm flexion movement times for combined test days, sexes, and trials were greatest for load 0 (41.7) and identical for load 1 (31.6) and load 2 (31.6).

<u>Interaction: loads by sex</u>. With 2 and 28 degrees of freedom an F of 5.49 is necessary for significance at the .01 level. The observed F of 7.05 exceeded the value necessary for significance. Thus, maximum speed forearm flexion movement times over loads did not follow the same pattern for each sex level. As depicted in Figure 9, increases in maximum speed forearm flexion movement time across loads was greater for females than males. This was evident in the 22.2% increase in maximum speed forearm flexion movement time for females from load zero (158.1) to load two (203.1 ms) as compared to the 21.4% increase in maximum speed forearm flexion movement time for males from load zero (122.5 ms) to load two (155.8 ms). Although statistically significant, the less than one percent difference in increase of maximum speed forearm flexion movement time between males and females with increasing load is of little practical significance.

Standard deviations of maximum speed forearm flexion movement time measures for females were greater for each load level than males as shown in Table 4. The largest standard deviations were for load zero (49.2) for females compared to load zero (20.8) and load one (23.7) for males.

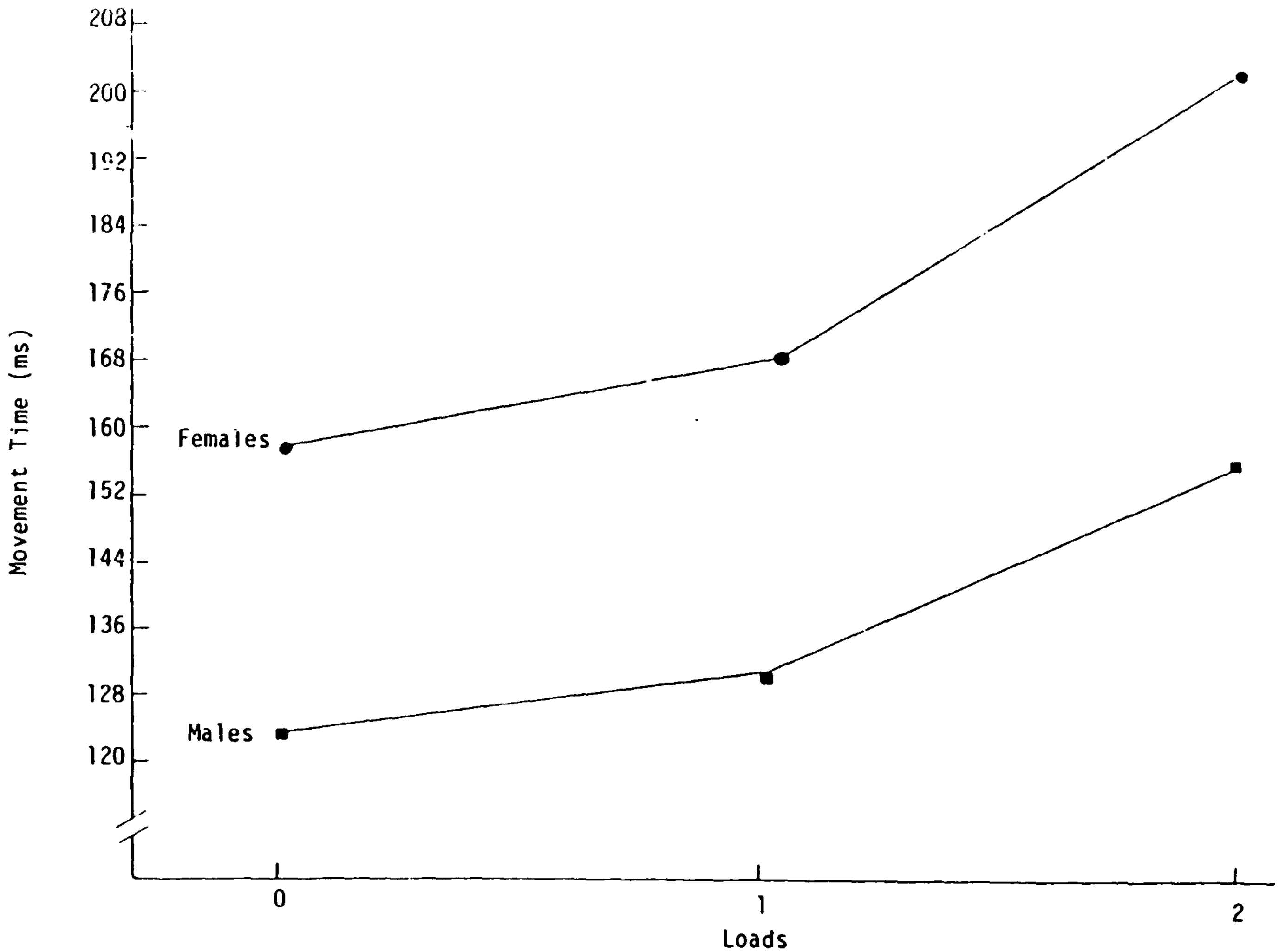


Figure 9. Maximum speed forearm flexion movement times (ms) across loads for females and m. e

Interaction: days by trials. With 18 and 252 degrees of freedom, an F of 1.62 is required for significance at the .05 level. The observed F of 1.75 exceeded this value necessary for significance. Thus, the pattern of maximum speed forearm flexion movement times over trials was not the same for each test day level. Decreases in maximum speed forearm flexion movement times over trials were greater on test day one than test days four and eight as shown in Table 5 and depicted in Figure 10. This finding is particularly striking since it suggests that rapid trial to trial improvements in maximum speed forearm flexion movement times occur on the first test day of practice with subsequent improvements occuring from day to day rather than trial to trial.

Standard deviations of maximum speed forearm flexion movement times were greater for trials one (98.7), two (40.0), and three (49.5) on test day one compared to standard deviations for all other trials on test days one, four, and eight for combined loads and sexes as shown in Table 5. After trials one, two, and three on test day one, the standard deviations of maximum speed forearm flexion movement time were generally similar.

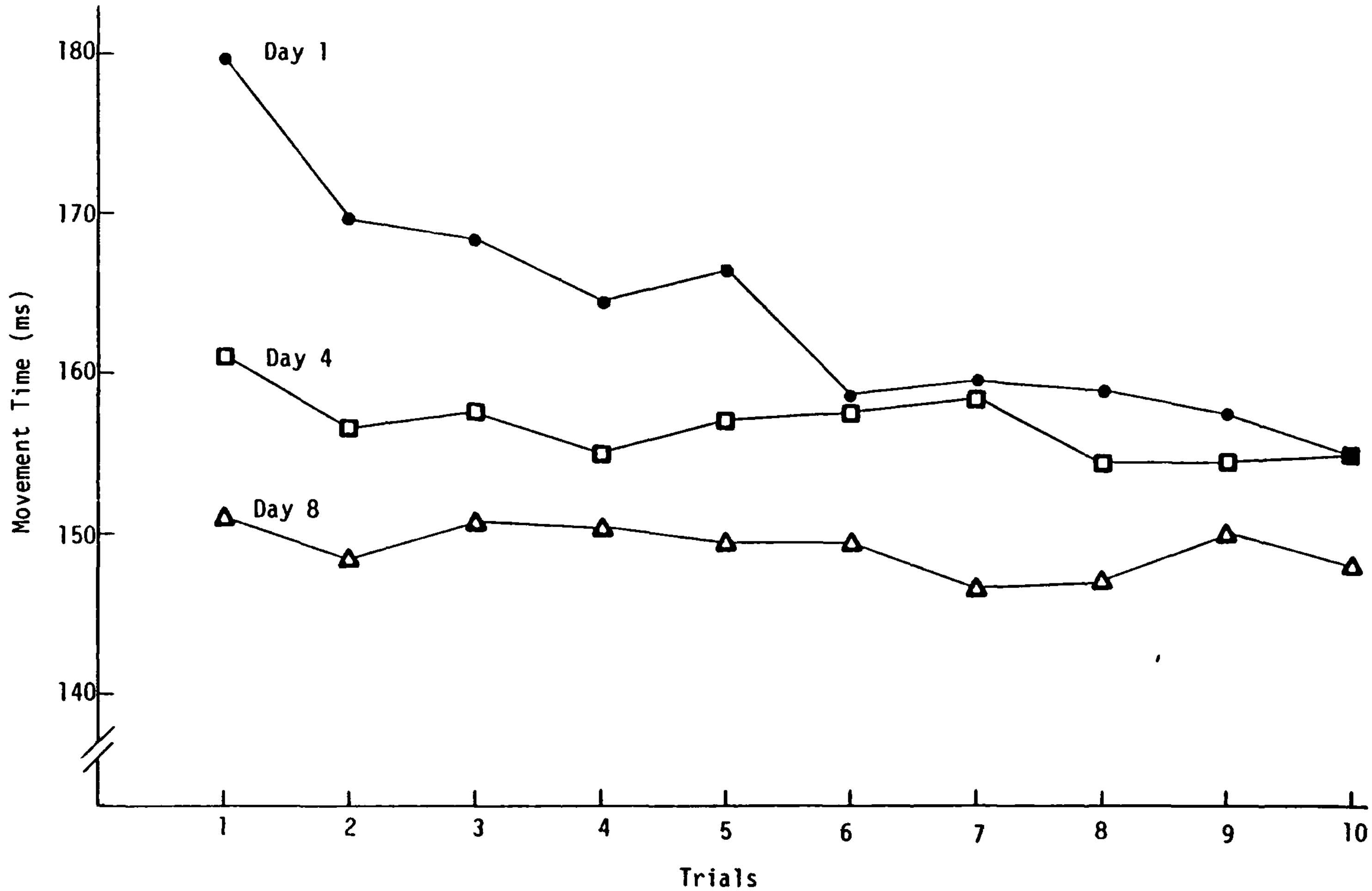


Figure 10. Maximum speed forearm flexion movement times (ms) across trials on test days one, four, and eight.

Maximal speed forearm flexion percent acceleration time. Means and standard deviations of maximum speed forearm flexion percent acceleration times for test days 1, 4, and 8 and combined days are shown in Table 8. Table 8 includes means and standard deviations for maximum speed forearm flexion percent acceleration times for males and females, loads, zero, one, and two, as well as for combined sexes, combined loads, and combined trials. Means and standard deviations for trial one through trial ten for combined test days, sexes, and loads are presented in Table 9. The four-way factorial analysis of variance of maximum speed forearm flexion percent acceleration time for days, sexes, loads, and trials is presented in Table 10.

Main effect: test days 1, 4, 8. Maximum speed forearm flexion percent acceleration times for combined sexes, loads, and trials increased over test days one (79.6%), four (83.2%) and eight (84.8%). With 2 and 28 degrees of freedom, an F of 3.34 is required for significance at the .05 level. The observed F for between test days of .06 did not exceed the value necessary for significance. Therefore, no significant differences between test days for maximum speed forearm flexion percent acceleration time were observed. There was no evidence to suggest that the percentage of maximum speed forearm flexion movement time that is acceleration time (i.e. % acceleration time) increases with practice over days as was previously reported by Lagasse [47] and Wolcott [95] for male subjects. Similar findings have been

TABLE 8

Means and standard deviations of maximum speed forearm flexion percent acceleration times (%) under three loads and combined loads for females and males and combined sexes over test days one, four, and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	75.1	13.0	77.2	11.7	67.3	13.6	73.2	13.4
Day 4	86.4	12.0	80.1	9.6	71.8	12.9	77.8	11.8
Day 8	82.6	12.5	84.3	9.6	75.7	11.4	80.9	11.8
Combined days	79.7	12.4	80.6	10.7	71.6	13.1	77.3	12.7
Males								
Day 1	86.4	12.0	86.9	12.3	84.5	11.3	85.9	11.9
Day 4	89.4	5.9	90.3	6.9	86.1	11.0	88.6	8.3
Day 8	89.8	8.8	89.2	9.0	87.0	9.9	88.7	9.3
Combined Days	88.5	9.3	88.8	9.7	85.8	10.8	87.7	10.0
Combined Sexes								
Day 1	80.8	13.7	82.1	12.9	75.9	15.2	79.6	14.2
Day 4	85.4	9.3	85.2	9.8	78.0	13.9	83.2	11.6
Day 8	86.2	11.4	86.8	9.6	81.3	12.0	84.8	11.3
Combined Days	84.1	11.8	84.7	11.0	78.7	13.9		

TABLE 9

Means and standard deviations for maximum speed forearm flexion percent acceleration time (%) across trials for combined test days, loads, and sexes.

	TRIALS									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	80.6	81.4	82.0	82.5	82.1	83.5	83.2	83.7	83.0	83.0
s.d.	14.3	12.7	12.6	12.3	12.9	11.8	11.4	13.1	12.0	12.6

TABLE 10

Analysis of variance for maximum speed forearm flexion percent acceleration time

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	9803932.26929	7185.15
SEX	1	35694.55068	26.16**
ERROR	14	1364.47140	
DAYS	2	114.54152	.06
DS	2	1071.98756	.61
ERROR	28	1762.75140	
LOAD	2	3402.91652	12.44**
LS	2	187.02188	.68
ERROR	28	273.46082	
DL	4	32.77547	.07
DLS	4	161.53187	.33
ERROR	56	483.49043	
TRIAL	9	141.87398	1.77
TS	9	39.13279	.49
ERROR	126	80.09349	
DT	18	18.07777	.31
DTS	18	64.56099	1.10
ERROR	252	58.71335	
LT	18	62.53573	1.02
LTS	18	64.71768	1.05
ERROR	252	61.41574	
DLT	36	67.13259	1.16
DLTS	36	48.40157	.84
ERROR	504	57.65611	

* Significant at the .05 level

** Significant at the .01 level

reported ,however, by Teves [81] for female subjects.

<u>Main effect: trials</u>. Maximal speed forearm flexion percent acceleration times were similar across trials as shown in Table 9. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 1.77 failed to exceed this level. Thus, no significant differences were observed between trials for maximum speed forearm flexion percent acceleration time. The results of similar comparisons have not been previously reported.

<u>Main effect: sex</u>. Maximum speed forearm flexion percent acceleration times for combined test days, loads, and trials were greater for males (87.7%) than females (77.3%). With 1 and 14 degrees of freedom, an F of 8.86 is required for significance at the .01 level. The observed F of 26.16 for the sex main effect exceeded this value necessary for significance. Thus, maximum speed forearm flexion percent acceleration times were significantly different between the sexes. This finding has not previously been reported. Standard deviations of maximum speed forearm flexion percent acceleration times for combined days, loads, and trials were slightly larger for females (12.7) than males (10.0).

<u>Main effect: loads</u>. Maximum speed forearm flexion percent acceleration times for combined test days, sexes, and trials were similar for load zero (84.1%) and load one (84.7%) but lower for load two (78.7%). With 2 and 28 degrees of freedom an F of 5.49

is required for significance at the .01 level. The observed F for between loads of 12.44 exceeded the value necessary for significance. Maximum speed forearm flexion percent acceleration times were therefore significantly different between loads. The results of mean separation procedures [51] revealed that maximum speed forearm flexion percent acceleration times were not significantly different between load zero and load one while both were significantly different from load two. Similar comparisons between loads have not previously been reported. Standard deviations of maximum speed forearm flexion movement times for combined test days, sexes, and trials were slightly greater for load two (13.9) than load zero (11.8) and load one (11.0).

Analysis of Variance of EMG Spike Parameters

Number of spikes for biceps brachii motor time digital raw EMG signals. Means and standard deviations of number of spikes for biceps brachii motor time digital raw EMG signals for test days 1, 4, and 8 and combined days are presented in Table 11. Also contained in Table 11 are means and standard deviations of number of spikes for biceps brachii motor time digital raw EMG signals for females and males and combined sexes as well as loads zero, one, two and combined loads for combined trials. Means and

TABLE 11

Means and standard deviations of number of spikes for biceps motor time digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	8.1	1.6	8.4	2.0	8.5	1.9	8.3	1.8
Day 4	7.1	1.8	7.1	1.9	7.9	2.0	7.4	1.9
Day 8	8.0	2.2	8.7	2.0	9.3	1.9	8.7	2.1
Combined Days	7.7	1.9	8.1	2.1	8.6	2.0	8.1	2.0
Males								
Day 1	6.3	2.3	6.9	2.7	8.2	2.7	7.1	2.7
Day 4	7.2	2.1	7.5	2.6	8.5	2.5	7.7	2.5
Day 8	6.7	2.1	7.4	2.0	8.9	2.7	7.7	2.4
Combined Days	6.7	2.2	7.3	2.4	8.5	2.6	7.5	2.5
Combined Sexes								
Day 1	7.2	2.2	7.6	2.5	8.3	2.3	7.7	2.4
Day 4	7.2	1.9	7.3	2.3	8.2	2.3	7.6	2.2
Day 8	6.6	3.4	6.8	3.4	8.2	3.3	8.2	2.3
Combined Days	7.2	2.1	7.7	2.3	8.5	2.3		

TABLE 12

Means and standard deviations for number of spikes for biceps brachii motor time digital raw EMG signals across trials for combined test days, loads, and sexes.

	TRIALS									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	7.9	7.7	7.6	8.0	7.7	7.8	7.6	7.8	7.9	7.9
s.d.	2.2	2.3	2.2	2.3	2.4	2.2	2.3	2.3	2.4	2.5

TABLE 13

Analysis of variance for number of spikes for biceps brachii motor time digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	87894.09378	472.53
SEX	1	135.80576	.73
ERROR	14	186.00909	
DAYS	2	49.00458	3.04
DS	2	84.76234	5.26*
ERROR	28	16.11508	
LOAD	2	215.24081	24.04**
LS	2	32.64014	3.65*
ERROR	28	8.95247	
DL	4	6.64367	1.66
DLS	4	2.93032	.73
ERROR	56	4.00957	
TRIAL	9	2.75277	1.06
TS	9	2.05195	.79
ERROR	126	2.59018	
DT	18	1.99094	.93
DTS	18	3.11028	1.45
ERROR	252	2.15060	
LT	18	4.19440	1.54
LTS	18	2.34995	.86
ERROR	252	2.73047	
DLT	36	3.07465	1.26
DLTS	36	2.84381	1.17
ERROR	504	2.43499	

* Significant at the .05 level

** Significant at the .01 level

standard deviations for trials one through ten for combined test days, loads, and sexes are displayed in Table 12. The four factor analysis of variance of number of spikes for biceps brachii motor time digital raw EMG signals for days sexes, loads and trials main effects is presented in Table 13.

Main effect: test days 1, 4, 8. Numbers of spikes for biceps brachii motor time digital raw EMG signals for combined sexes, loads, and trials increased slightly from test days one (7.7 spikes) and four (7.6 spikes) to test day eight (8.2 spikes). With 2 and 28 degrees of freedom, an F of 3.34 is required for significance at the .05 level. The observed F of 3.04 approached but did not exceed this required value. Thus, there were no significant differences between number of spikes for biceps brachii motor time digital raw EMG signals over the three test days. There was no evidence suggesting that number of spikes for biceps brachii motor time digital raw EMG signals increases with practice over days.

Main effect: trials. Number of spikes for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and loads, shown in Table 12, were similar across all trials. With 9 and 126 degrees of freedom, an F of 1.95 is required for significance at the .05 level. The observed F of 1.06 failed to surpass this value necessary for significance. Thus, no significant differences existed between trials for number of spikes for biceps brachii motor time digital raw EMG signals.

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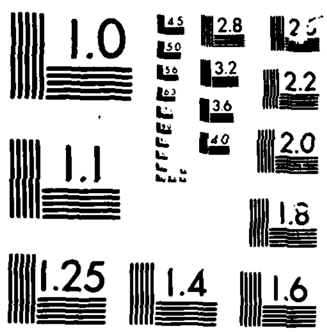

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Number of spikes for biceps brachii motor time digital raw EMG signals did not change across trials.

Main effect: sex. As seen in Table 11, number of spikes for biceps brachii motor time digital raw EMG signals for combined test days, loads, and trials were slightly more for females (8.1 spikes) than males (7.5 spikes). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.73 for between females and males failed to surpass this required value for significance. Hence, number of spikes for biceps brachii motor time digital raw EMG signals were not significantly different between males and females. The number of spikes for biceps brachii motor time digital raw EMG signals were not greater for males than females.

Main effect: loads. As seen in Table 11, number of spikes for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and trials increased as load increased from load zero (7.2 spikes) to load one (7.7 spikes) to load two (8.5 spikes). With 2 and 28 degrees of freedom an F of 5.45 is required for significance at the .01 level. The observed F for between loads zero, one, and two of 24.04 exceeded the value necessary for significance. Consequently, significant differences existed between loads for number of spikes for biceps brachii motor time digital raw EMG signals. Mean separation procedures revealed that number of spikes for biceps brachii motor time digital raw EMG signals were not significantly

different between loads zero and one while these loads were significantly different from load two. Thus, number of spikes for biceps brachii motor time digital raw EMG signals were greater for load two than loads zero and one.

Interaction: days by sex. With 2 and 28 degrees of freedom an F of 3.34 is necessary for significance at the .05 level. The observed F of 5.26 exceeded this value required for significance. Hence, number of spikes for biceps brachii motor time digital raw EMG signals over test days did not follow the same pattern for each sex level. As shown in Figure 11, number of spikes for biceps brachii motor time digital raw EMG signals increased for males and decreased for females between test days one and four. Males stabilized between days four and eight while females increased number of spikes for biceps brachii motor time digital raw EMG signals. Standard deviations were slightly greater for males than females as shown in Table 11.

Interaction: loads by sex. With 2 and 28 degrees of freedom an F of 3.34 is required for significance at the .01 level. The observed F of 3.65 surpassed this value necessary for significance. Therefore, the pattern of number of spikes for biceps brachii motor time digital raw EMG signals over loads was different for each sex level. As graphically represented in Figure 12, the number of spikes for biceps brachii motor time digital raw EMG signals increased over loads for both females and males. Males increased to a greater extent under load two than

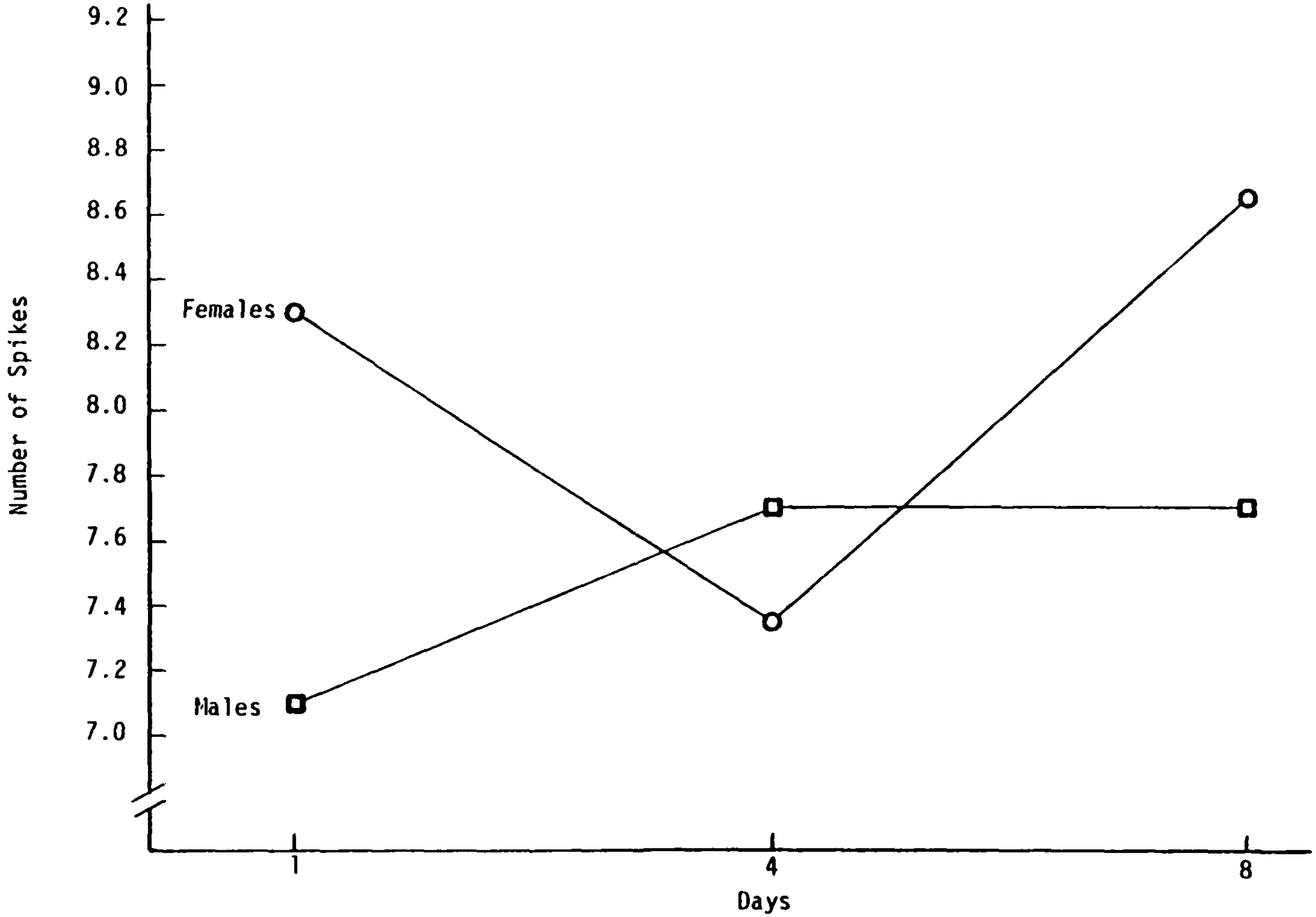

Figure 11. Number of spikes for biceps brachii motor time digital raw EMG signals across test days for females and males.

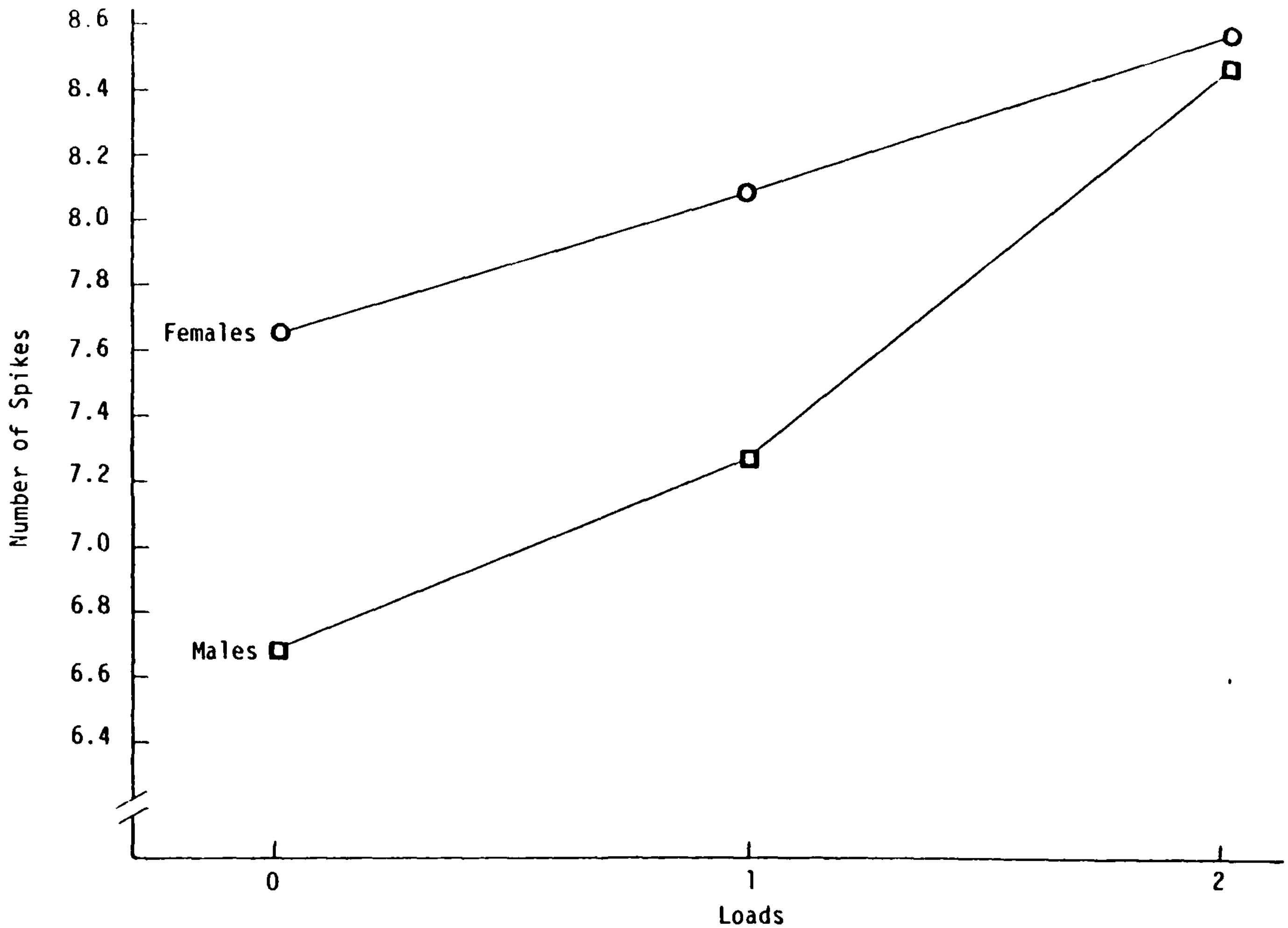

Figure 12. Number of spikes for biceps brachii motor time digital raw EMG signals across loads for females and males.

did females, resulting in similar number of spikes for biceps brachii motor time digital raw EMG signals shown in Figure 12. Heavier loads (load 2) seem ,therefore, to have an equalizing effect on the number of spikes for biceps brachii motor time digital raw EMG signals between males and females.

Mean spike amplitude for biceps brachii motor time digital raw EMG signals. Means and standard deviations of mean spike amplitude for biceps brachii motor time digital raw EMG signals for test days one, four and eight and combined days are shown in Table 14. Included in Table 14 are mean spike amplitude for biceps brachii motor time means and standard deviations for males and females, loads zero, one, and two, as well as for combined sexes, combined loads and combined trials. Means and standard deviations for trials one through ten are shown in Table 15 for combined days, sexes, and loads. The four-factor analysis of variance of mean spike amplitude for biceps brachii motor time digital raw EMG signals for days, sexes, loads, and trials is presented in Table 16.

Main effect: test days 1, 4, 8. Mean spike amplitude for biceps brachii motor time digital raw EMG signals for combined sexes, loads and trials were slightly lower on test day four (1.046) than on test days one (1.192 mv) and eight (1.195 mv). With 2 and 28 degrees of freedom an F of 3.34 is required for

TABLE 14

Means and standard deviations of mean spike amplitude (mv) for biceps brachii motor time digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	.853	.498	.985	.822	1.048	.813	.962	.728
Day 4	.938	.612	1.003	.753	1.046	.861	.996	.747
Day 8	1.398	1.687	.977	.784	1.076	.788	1.150	1.175
Combined days	1.063	1.097	.989	.784	1.057	.818	1.036	.910
Males								
Day 1	.995	1.275	1.759	2.238	1.511	2.246	1.422	1.991
Day 4	1.085	.500	1.142	.600	1.059	.500	1.095	.557
Day 8	1.212	.926	1.208	.784	1.297	.831	1.239	.846
Combined Days	1.098	.955	1.370	1.442	1.289	1.418	1.252	1.295
Combined Sexes								
Day 1	.924	.967	1.372	1.724	1.280	1.699	1.192	1.515
Day 4	1.012	.562	1.073	.709	1.053	.702	1.046	.661
Day 8	1.305	1.360	1.093	.790	1.187	.814	1.195	1.024
Combined days	1.080	1.027	1.179	1.175	1.173	1.162		

TABLE 15

Means and standard deviations for mean spike amplitude (mv) for biceps brachii motor time digital raw EMG signals across trials for combined test days, loads, and sexes.

	TRIALS									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	1.149	1.101	1.211	1.098	1.153	1.042	1.163	1.166	1.120	1.239
s.d.	1.154	1.027	1.261	1.021	1.185	.870	1.183	1.136	1.075	1.284

TABLE 16

Analysis of variance for mean spike amplitude for biceps brachii motor time digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	1884.88942	44.82
SEX	1	16.80353	.40
ERROR	14	42.05378	
DAYS	2	3.49509	.18
DS	2	5.34503	.27
ERROR	28	19.59971	
LOAD	2	1.47331	.33
LS	2	3.62809	.81
ERROR	28	4.48857	
DL	4	4.72713	1.00
DLS	4	1.42141	.30
ERROR	56	4.75038	
TRIAL	9	.47251	2.46*
TS	9	.20661	1.08
ERROR	126	.19182	
DT	18	.17495	.99
DTS	18	.18711	1.06
ERROR	252	.17677	
LT	18	.06555	.42
LTS	18	.21549	1.38
ERROR	252	.15586	
DLT	36	.18306	1.14
DLTS	36	.14488	.91
ERROR	504	.16008	

* Significant at the .05 level

** Significant at the .01 level

significance at the .05 level. The observed F of 0.18 failed to exceed this value necessary for significance. Hence, there were no significant differences between mean spike amplitude for biceps brachii motor time digital raw EMG signals over the three test days. Practice over days had no discernible effect on mean spike amplitude for biceps brachii motor time digital raw EMG signals.

<u>Main effect: trials</u>. Means for mean spike amplitude for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and loads presented in Table 15, were greatest for trial six (1.C42 mv) and least for trial ten (1.239 mv). With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 2.46 exceeded this value necessary for significance. Hence, significant differences existed between mean spike amplitude for biceps brachii motor time digital raw EMG signals means for trials one through ten. The results of mean separation procedures [51] are presented in Table 17. No identifiable pattern was discernible for mean spike amplitude for biceps brachii motor time digital raw EMG signals across trials from these results.

<u>Main effect: sex</u>. Mean spike amplitude for biceps brachii motor time digital raw EMG signals for combined test days, loads and trials were greater for males (1.252 mv) than females (1.036 mv). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.40 failed

TABLE 17

Lindquist's critical difference test applied to trial mean differences for combined test days, sexes, and loads for mean spike amplitude for biceps brachii motor time digital raw EMG signals.

Trials (ranked according to mean value)

6	4	2	9	1	5	7	8	3	10

to surpass this value necessary for significance. Thus, no significant differences existed between males and females. Whether a subject was male or female had no discernible influence on mean spike amplitude for biceps brachii motor time digital raw EMG signals.

Main effect: loads. As seen in Table 14, mean spike amplitude for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and trials were slightly lower for load zero (1.080 mv) than loads one (1.179 mv) and two (1.173). With 2 and 28 degrees of freedom an F of 3.34 is required for significance at the .05 level. The observed F for between loads of 1.47 did not exceed this value necessary for significance. Mean spike amplitude for biceps brachii motor time digital raw EMG signals were therefore not significantly different between load zero, one, and two. Mean spike amplitude for biceps brachii motor time digital raw EMG signals was not discernibly lower for load zero than loads one and two.

Mean spike duration for biceps brachii motor time digital raw EMG signals. Means and standard deviations of mean spike duration for biceps brachii motor time digital raw EMG signals for test days one, four, and eight and combined days are presented in Table 18. Also in Table 18 are mean spike duration for biceps brachii motor time digital raw EMG signals means and standard

TABLE 18

Means and standard deviations of mean spike duration (ms) for biceps brachii motor time digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	12.0	2.5	12.2	2.7	12.6	2.4	12.2	2.6
Day 4	11.3	2.5	11.7	2.3	11.7	2.2	11.6	2.3
Day 8	10.6	2.2	11.2	2.5	10.8	2.1	10.9	2.3
Combined Days	11.3	2.5	11.7	2.6	11.7	2.3	11.6	2.5
Males								
Day 1	12.4	3.5	13.1	5.2	12.5	3.9	12.7	4.2
Day 4	10.2	2.0	11.3	2.7	11.1	2.3	10.9	2.4
Day 8	10.9	2.6	11.0	2.4	10.7	2.5	10.9	2.5
Combined Days	11.2	2.9	11.9	3.7	11.4	3.1	11.5	3.3
Combined Sexes								
Day 1	12.2	3.0	12.6	4.1	12.5	3.2	12.5	3.5
Day 4	10.8	2.3	11.5	2.4	11.4	2.9	11.2	2.4
Day 8	10.7	2.4	11.1	2.4	10.8	2.3	10.9	2.4
Combined Days	11.2	2.7	11.7	3.2	11.6	2.7		

deviations for males and females, loads zero, one and two, as well as for combined sexes, combined loads and combined trials. Means and standard deviations for trials one through ten for combined days, sexes, and loads are shown in Table 19. The four-factor analysis of variance of mean spike duration for biceps brachii motor time digital raw EMG signals for days, sexes, loads, and trials is presented in Table 20.

Main effect: test days 1, 4, 8. Mean spike duration for biceps brachii motor time digital raw EMG signals for combined sexes, loads, and trials were lower on test days four (11.2 ms) and eight (10.9 ms) than on test day one (12.5 ms). With 2 and 28 degrees of freedom an F of 5.45 is required for significance at the .01 level. The observed F of 10.35 exceeded this value necessary for significance. Thus, significant differences existed for mean spike duration for biceps brachii motor time digital raw EMG signals between test day levels. Mean separation procedures [51] revealed that mean spike duration for biceps brachii motor time digital raw EMG signals for test days four and eight were not significantly different while mean spike duration for biceps brachii motor time digital raw EMG signals for test day one was significantly different from test days four and eight. Mean spike duration for biceps brachii motor time digital raw EMG signals decreased between pre-practice test day one and post-practice test days four and eight. Standard deviations also decreased from test day one (3.5) to test days

TABLE 19

Means and standard deviations for mean spike duration (ms) across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	11.4	11.5	11.6	11.3	11.5	11.5	11.8	11.5	11.6	11.5
s.d.	2.8	2.6	2.8	2.7	3.7	2.7	3.1	2.6	3.1	2.8

TABLE 20

Analysis of variance for mean spike duration for biceps brachii motor time digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	190955.91925	757.87
SEX	1	3.06946	.01
ERROR	14	251.96413	
DAYS	2	331.77314	10.35**
DS	2	39.78262	1.24
ERROR	28	32.06338	
LOAD	2	31.99037	2.51
LS	2	5.28784	.41
ERROR	28	12.76167	
DL	4	3.76405	.41
DLS	4	5.56955	.61
ERROR	56	9.18048	
TRIAL	9	2.76788	.64
TS	9	6.24919	1.43
ERROR	126	4.35865	
DT	18	3.91313	.92
DTS	18	6.94266	1.63
ERROR	252	4.26892	
LT	18	4.54148	.97
LTS	18	3.18733	.68
ERROR	252	4.68832	
DLT	36	5.72172	1.26
DLTS	36	4.27550	.94
ERROR	504	4.53787	

* Significant at the .05 level

** Significant at the .01 level

four (2.4) and eight (2.4).

Main effect: trials. Means for mean spike duration for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and loads presented in Table 19, were similar across all ten trials. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 0.64 failed to exceed this value necessary for significance. Thus, there were no significant differences between trials for mean spike duration for biceps brachii motor time digital raw EMG signals.

Main effect: sex. As seen in Table 18, mean spike duration for biceps brachii motor time digital raw EMG signals for combined test days, loads and trials were very similar for males (11.5 ms) and females (11.6 ms). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.01 failed to exceed this value required for significance. Hence, there were no significant differences between sexes for mean spike duration for biceps brachii motor time digital raw EMG signals.

Main effect: loads. As seen in Table 18, mean spike duration for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and trials were slightly less for load zero (11.2 ms) than loads one (11.7 ms) and two (11.6 ms). With 2 and 28 degrees of freedom an F of 3.34 is required for significance at the .05 level. The observed F of 2.51 failed to

surpass this value necessary for significance. No significant differences, therefore, existed for mean spike duration for biceps brachii motor time digital raw EMG signals between loads.

Mean number of peaks per spike for biceps brachii motor time digital raw EMG signals. Means and standard deviations of mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for test days one, four, and eight and combined days are presented in Table 21. Included in Table 21 are means and standard deviations of mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for females and males and combined sexes as well as loads zero, one, and two and combined loads for combined trials. Means and standard deviations for trial one through ten for males and females and combined test days and loads are displayed in Table 22. The four-factor analysis of variance of mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for days, sexes, loads, and trials is presented in Table 23.

Main effect: test days 1, 4, 8. Mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for combined sexes, loads, and trials were similar over test days one (2.1 peaks), four (1.9 peaks) and eight (2.0 peaks). With 2 and 28 degrees of freedom, an F of 3.34 is required for significance

TABLE 21

Means and standard deviations of mean number of peaks per spike for biceps brachii motor time digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d	Mean	s.d.	Mean	s.d.
Day 1	2.0	0.4	2.0	0.5	2.1	0.5	2.0	0.5
Day 4	2.0	0.4	2.0	0.4	2.0	0.4	2.0	0.4
Day 8	2.0	0.5	2.1	0.5	2.0	0.3	2.0	0.4
Combined Days	2.0	0.4	2.0	0.5	2.1	0.4	2.0	0.4
Males								
Day 1	2.1	0.7	2.2	1.0	2.0	0.5	2.1	0.7
Day 4	1.8	0.4	1.9	0.4	2.0	0.5	1.9	0.4
Day 8	2.0	0.6	2.0	0.5	2.0	0.5	2.0	0.5
Combined Days	2.0	0.6	2.0	0.7	2.0	0.5	2.0	0.6
Combined Sexes								
Day 1	2.0	0.6	2.1	0.8	2.1	0.5	2.1	0.6
Day 4	1.9	0.4	1.9	0.4	2.0	0.4	1.9	0.4
Day 8	2.0	0.5	2.0	0.5	2.0	0.4	2.0	0.5
Combined Days	2.0	0.5	2.0	0.6	2.0	0.5		

TABLE 22

Mean and standard deviations for mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for across trials and sexes and for combined sexes and for all combined test days and loads.

	Trials									
Females	1	2	3	4	5	6	7	8	9	10
Mean	2.1	2.0	2.0	1.9	2.0	2.0	2.1	2.0	2.0	2.1
s.d.	0.4	0.4	0.4	0.4	0.4	0.4	0.4	0.5	0.5	0.5
Males										
Mean	1.9	2.0	2.0	2.1	2.0	2.1	2.0	1.9	1.9	2.0
s.d.	0.5	0.6	0.5	0.5	0.9	0.6	0.5	0.5	0.6	0.6
Combined Sex										
Mean	2.0	2.0	2.0	2.0	2.0	2.1	2.0	2.0	2.0	2.0
s.d.	0.4	0.5	0.5	0.4	0.7	0.5	0.5	0.5	0.5	0.5

TABLE 23

Analysis of variance for mean number of peaks per spike for biceps brachii motor time digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	5774.63769	1524.73
SEX	1	.07954	.02
ERROR	14	3.78732	
DAYS	2	1.63160	2.07
DS	2	.89401	1.13
ERROR	28	.78837	
LOAD	2	.39690	.85
LS	2	.21113	.45
ERROR	28	.46762	
DL	4	.37109	1.10
DLS	4	.57313	1.70
ERROR	56	.33619	
TRIAL	9	.17009	.99
TS	9	.39146	2.28*
ERROR	126	.17186	
DT	18	.22093	.98
DTS	18	.20242	.90
ERROR	252	.22532	
LT	18	.26765	1.25
LTS	18	.13714	.64
ERROR	252	.21464	
DLT	36	.20816	1.03
DLTS	36	.18247	.90
ERROR	504	.20167	

* Significant at the .05 level

** Significant at the .01 level

at the .05 level. The observed F of 2.07 failed to exceed this value necessary for significance. Thus, there were no significant differences between mean number of peaks per spike for biceps brachii motor time digital raw EMG signals over the three test days. There was no evidence suggesting that mean number of peaks per spike for biceps brachii motor time digital raw EMG signals changes with practice over days.

Main effect: trials. Mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and loads shown in Table 22, were the same for all trials (2.0 peaks) except trial six (2.1 peaks). With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 0.99 did not exceed this value necessary for significance. Thus, no significant differences existed for mean number of peaks per spike for biceps brachii motor time digital raw EMG signals between trials.

Main effect: sex. As seen in Table 21, mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for combined test days, loads, and trials were the same for females (2.0 peaks) and males (2.0 peaks). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.02 failed to exceed this value necessary for significance. No significant differences existed between females and males for mean number of peaks per spike for biceps brachii motor time digital raw EMG signals.

Main effect: loads. As seen in Table 21, mean number of peaks per spike for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and trials were the same for load zero (2.0 peaks), one (2.0 peaks) and two (2.0 peaks). With 2 and 28 degrees of freedom an F of 3.34 is required for significance at the .05 level. The observed F of 0.85 failed to surpass this value necessary for significance. Thus, mean number of peaks per spike for biceps brachii motor time digital raw EMG signals were not significantly different between loads.

Interaction: trial by sex. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 2.28 exceeded this value necessary for significance. Thus, mean number of peaks per spike for biceps brachii motor time digital raw EMG signals over trials did not follow the same pattern for females and males. Figure 13 shows mean number of peaks per spike for biceps brachii motor time digital raw EMG signals plotted across trials for females and males. It can be seen in Figure 13, that the pattern for males and females were different until trial seven while the pattern of subsequent trials were similar.

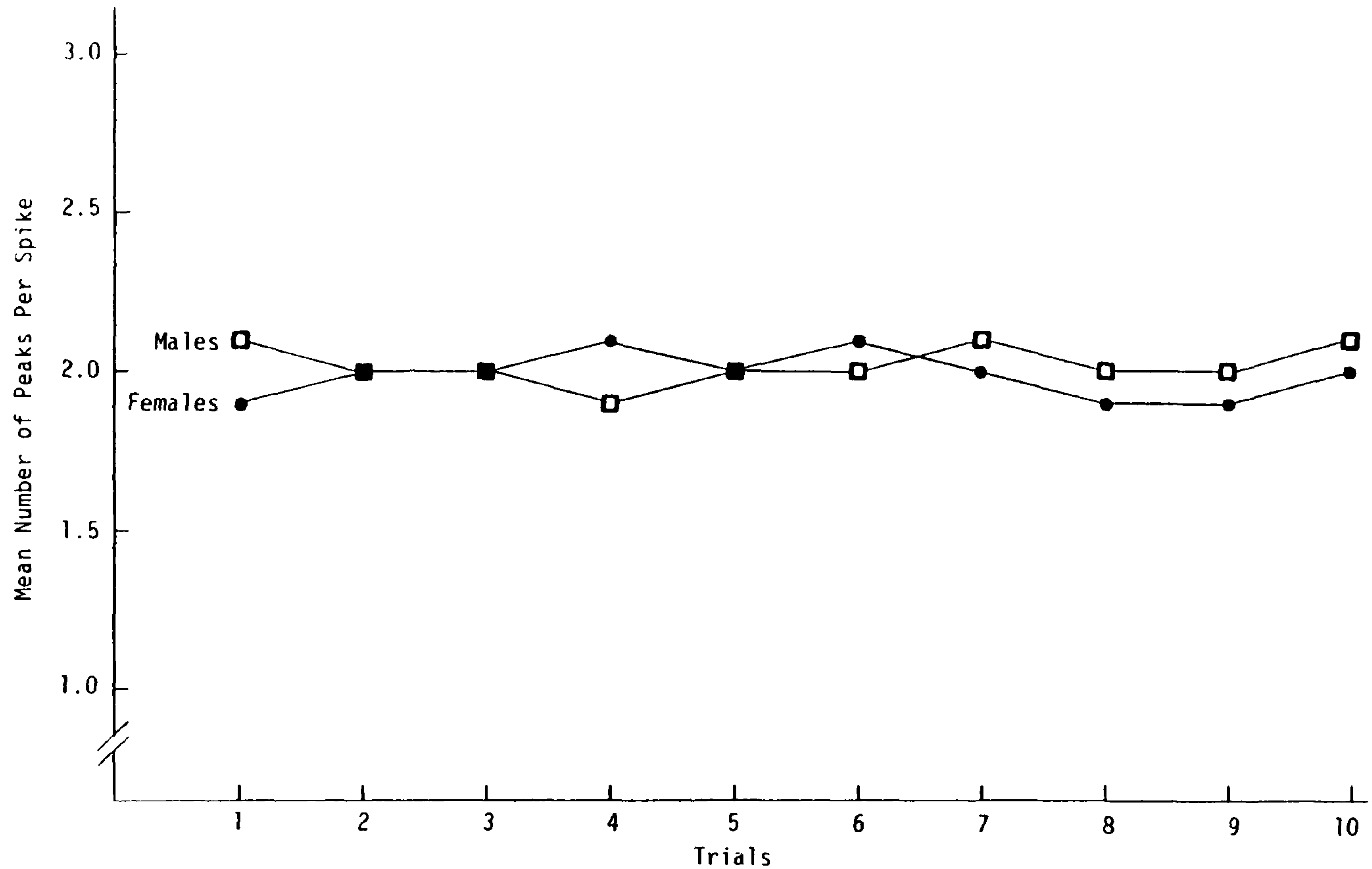

Figure 13. Mean number of peaks per spike for biceps brachii motor time digital raw EMG signals across trials for females and males.

Spike frequency for biceps brachii motor time digital raw EMG signals. Means and standard deviations of spike frequency for biceps brachii motor time digital raw EMG signals for test days one, four, and eight and combined days are presented in Table 24. Contained in Table 24 are means and standard deviations of spike frequency for biceps brachii motor time digital raw EMG signals for females and males and combined sexes as well as loads zero, one, and two and combined loads for combined trials. Means and standard deviations for trials one through ten for combined test days, loads, and sexes are presented in Table 25. The four-factor analysis of variance of spike frequency for biceps brachii motor time digital raw EMG signals for day, sex, load, and trial main effects is presented in Table 26.

Main effect: test days 1, 4, 8. Spike frequency for biceps brachii motor time digital raw EMG signals for combined sexes, loads, and trials increased from test day one (85.9 Hz) to test day four (93.2 Hz) to test day eight (96.5 Hz). With 2 and 28 degrees of freedom an F of 5.49 is required for significance at the .01 level. The observed F of 9.68 exceeded this value necessary for significance. Hence, significant differences existed between test day levels for spike frequency for biceps brachii motor time digital raw EMG signals. Mean separation procedures [51] revealed that test days four and eight were significantly different from test day one. Spike frequency for

TABLE 24

Means and standard deviations of spike frequency (Hz) for biceps brachii motor time digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	87.3	18.6	86.5	20.8	82.6	16.0	85.5	18.6
Day 4	92.1	18.8	88.9	17.9	88.4	16.3	89.8	17.7
Day 8	98.2	20.1	94.3	21.3	96.0	19.3	96.2	20.2
Combined Days	92.6	19.6	89.9	20.2	89.0	18.0	90.5	19.3
Males								
Day 1	86.7	22.8	85.0	17.8	87.1	24.8	86.2	24.3
Day 4	101.8	21.4	94.3	21.3	94.4	21.0	96.6	22.0
Day 8	97.1	22.3	95.0	19.3	98.6	22.2	96.7	21.3
Combined Days	95.2	23.0	91.0	23.0	93.4	23.1	93.2	23.1
Combined Sexes								
Day 1	87.0	20.7	85.7	23.2	84.8	20.9	85.9	21.6
Day 4	96.9	20.7	91.3	20.6	91.4	18.9	93.2	20.2
Day 8	97.6	21.2	94.4	20.3	97.3	20.8	96.5	20.8
Combined days	93.9	21.4	90.5	21.7	91.2	20.8		

TABLE 25

Means and standard deviations for spike frequency (Hz) for biceps motor time digital raw EMG signals across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	92.7	91.8	91.4	93.0	92.2	91.5	89.5	91.7	92.3	92.3
s.d.	22.0	21.2	21.3	20.8	20.0	20.5	21.0	21.2	23.1	22.4

TABLE 26

Analysis of variance for spike frequency for biceps brachii motor time digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	12145709.82616	809.77
SEX	1	2639.47544	.18
ERROR	14	14998.88355	
DAYS	2	14171.30286	9.68**
DS	2	1493.44974	1.02
ERROR	28	1463.59615	
LOAD	2	1537.30892	2.53
LS	2	306.75902	.50
ERROR	28	608.10419	
DL	4	415.59187	1.05
DLS	4	261.12251	.66
ERROR	56	396.24258	
TRIAL	9	136.26499	.56
TS	9	266.17668	1.09
ERROR	126	245.29923	
DT	18	224.10708	.92
DTS	18	347.32757	1.43
ERROR	252	243.18332	
LT	18	351.94357	1.25
LTS	18	170.75084	.60
ERROR	252	282.59944	
DLT	36	288.53827	1.23
DLTS	36	237.10653	1.01
ERROR	504	234.65685	

* Significant at the .05 level

** Significant at the .01 level

biceps brachii motor time digital raw EMG signals increased from test day one to test days four and eight. Standard deviations were similar on test days one (21.6), four (20.2) and eight (20.8).

Main effect: trials. Means for spike frequency for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and loads presented in Table 25, were least on trial seven (89.5 Hz) and greatest on trial four (93.0 Hz). With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 0.56 failed to exceed this value necessary for significance. Hence, no significant differences existed between trials for spike frequency for biceps brachii motor time digital raw EMG signals.

Main effect: sex. As seen in Table 24, spike frequency for biceps brachii motor time digital raw EMG signals for combined test days, loads, and trials were slightly greater for males (93.2 Hz) than females (90.5 Hz). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.18 failed to surpass this value necessary for significance. Thus, there were no significant differences between sexes for spike frequency for biceps brachii motor time digital raw EMG signals.

Main effect: loads. As seen in Table 24, spike frequency for biceps brachii motor time digital raw EMG signals for combined test days, sexes, and trials were slightly greater for

load zero (93.9 Hz) than loads one (90.5 Hz) and two (91.2 Hz). With 2 and 28 degrees of freedom an F of 3.34 is required for significance at the .05 level. The observed F of 2.53 failed to exceed this value necessary for significance. Therefore, no significant differences existed between loads for spike frequency for biceps brachii motor time digital raw EMG signals.

Number of spikes for end of the first biceps brachii burst digital raw EMG signals. Means and standard deviations of number of spikes for end of the first biceps brachii burst digital raw EMG signals for test days one, four, and eight and combined days are shown in Table 27. Also shown in Table 27 are means and standard deviations of number of spikes for end of the first biceps brachii burst digital raw EMG signals for males and females, loads zero, one and two, as well as for combined sexes, combined loads and combined trials. Means and standard deviations for trial one through trial ten for combined test days, sexes, and loads are presented in Table 28. The four-way factorial analysis of variance of number of spikes for end of the first biceps brachii burst digital raw EMG signals for days, sexes, loads, and trials is presented in Table 29.

Main effect: test days 1, 4, 8. Number of spikes for end of the first biceps brachii burst digital raw EMG signals for combined sexes, loads, and trials were greater for test day one

TABLE 27

Means and standard deviations of number of spikes for end of the first biceps brachii burst digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	10.0	5.6	9.0	4.4	10.3	4.2	9.7	4.8
Day 4	6.5	2.1	8.1	3.6	9.1	3.2	7.9	3.2
Day 8	8.0	3.8	8.2	3.6	10.0	3.3	8.7	3.7
Combined Days	8.2	4.3	8.4	3.9	9.8	3.6	8.8	4.0
Males								
Day 1	5.4	2.9	6.0	3.7	6.5	2.2	5.6	3.0
Day 4	5.1	2.0	4.7	1.6	6.1	1.5	5.3	1.8
Day 8	5.2	2.1	5.4	2.4	6.4	2.2	5.6	2.3
Combined Days	5.2	2.4	5.3	2.7	6.3	2.0	5.6	2.4
Combined Sexes								
Day 1	7.7	5.0	7.5	4.3	8.4	3.8	7.8	4.4
Day 4	5.9	2.2	6.4	3.3	7.6	2.9	6.6	2.9
Day 8	6.6	3.4	6.8	3.4	8.2	3.3	7.2	3.4
Combined Days	6.7	3.8	6.9	3.7	8.1	3.4		

TABLE 28

Means and standard deviations for number of spikes for end of the first biceps brachii burst digital raw EMG signals across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	7.0	7.2	7.2	7.0	7.4	7.3	7.2	7.2	7.2	7.4
s.d.	3.4	3.7	3.8	3.7	3.9	3.2	3.8	3.7	3.7	3.8

TABLE 29

Analysis of variance for number of spikes for end of the first biceps brachii burst digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	74779.25625	379.02
SEX	1	3600.50625	18.25**
ERROR	14	197.29712	
DAYS	2	187.65625	2.78
DS	2	40.87708	.61
ERROR	28	67.55397	
LOAD	2	264.50833	14.80**
LS	2	6.47500	.36
ERROR	28	17.87421	
DL	4	16.45833	1.15
DLS	4	35.00833	2.45
ERROR	56	14.28373	
TRIAL	9	2.26860	.34
TS	9	3.98156	.61
ERROR	126	6.57622	
DT	18	7.67091	1.06
DTS	18	8.28989	1.14
ERROR	252	7.24742	
LT	18	3.97901	.62
LTS	18	5.94568	.92
ERROR	252	6.43042	
DLT	36	7.91744	1.23
DLTS	36	6.52762	1.01
ERROR	504	6.46164	

* Significant at the .05 level

** Significant at the .01 level

(7.8 spikes) than test days four (6.6 spikes) and eight (7.2 spikes). With 2 and 28 degrees of freedom, an F of 3.34 is required for significance at the .05 level. The observed F for between test days of 2.78 failed to exceed this value necessary for significance. Thus, no significant differences between test days for number of spikes for end of the first biceps brachii burst digital raw EMG signals were observed. There was no evidence to suggest that number of spikes for end of the first biceps brachii burst digital raw EMG signals decrease over test days due to practice.

Main effect: trials. Number of spikes for end of the first biceps brachii burst digital raw EMG signals were similar across trials as shown in Table 28. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 0.34 failed to exceed this value necessary for significance. Thus, no significant differences were observed between trials for number of spikes for end of the first biceps brachii burst digital raw EMG signals.

Main effect: sex. As seen in Table 27, number of spikes for end of the first biceps brachii burst digital raw EMG signals for combined test days, loads, and trials were greater for females (8.8 spikes) than males (5.6 spikes). With 1 and 14 degrees of freedom an F of 8.86 is required for significance at the .01 level. The observed F of 18.25 exceeded this value necessary for significance. Therefore, number of spikes for end of the first

biceps brachii burst digital raw EMG signals were significantly different between males and females. Females had a greater number of spikes for end of the first biceps brachii burst digital raw EMG signals than males. Standard deviations were also larger for females (4.0) than males (2.4).

Main effect: loads. As seen in Table 27, number of spikes for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and trials were greatest under load two (8.1 spikes) and least under loads zero (7.6 spikes) and one (6.9 spikes). With 2 and 28 degrees of freedom an F of 5.49 is required for significance at the .01 level. The observed F for number of spikes for end of the first biceps brachii burst digital raw EMG signals of 14.80 surpassed this value necessary for significance. Number of spikes for end of the first biceps brachii burst digital raw EMG signals were significantly different between loads. The results of mean separation procedures [51] revealed that number of spikes for end of the first biceps brachii burst digital raw EMG signals were not significantly different between load zero and load one while both were significantly different from load two. Number of spikes for end of the first biceps brachii burst digital raw EMG signals were greater on load two than loads zero and one. Standard deviations for number of spikes for end of the first biceps brachii burst digital raw EMG signals were similar for loads zero (3.8), one (3.7), and two (3.4).

Mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals. Means and standard deviations of mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals for test days one, four, and eight and combined days are presented in Table 30. Also contained in Table 30 are means and standard deviations of mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals for females and males and combined sexes as well as loads zero, one, and two and combined loads for combined trials. Means and standard deviations for trials one through ten for combined test days, loads, and sexes are displayed in Table 31. The four-factor analysis of variance of mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals for day, sex, load, and trial main effects is presented in Table 32.

Main effect: test days 1, 4, 8. Mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals for combined sexes, loads, and trials increased slightly from test days one (1.071 mv) and four (1.049 mv) to test day eight (1.153 mv). With 2 and 28 degrees of freedom, an F of 3.34 is required for significance at the .05 level. The observed F of 0.12 failed to exceed this value necessary for significance. Thus, there were no significant differences between mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals on

TABLE 30

Means and standard deviations of mean spike amplitude (mv) for end of the first biceps brachii burst digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	0.902	0.669	0.943	0.645	1.123	0.989	0.990	0.786
Day 4	1.047	0.709	1.015	0.783	1.218	1.080	1.094	0.873
Day 8	1.611	2.099	1.007	0.743	1.060	0.841	1.226	1.395
Combined Days	1.187	1.365	0.988	0.724	1.134	0.973	1.103	1.056
Males								
Day 1	0.922	0.936	1.289	1.482	1.244	1.808	1.152	1.457
Day 4	1.022	0.386	1.000	0.592	0.990	0.557	1.004	0.518
Day 8	1.114	0.732	0.984	0.507	1.140	0.632	1.080	0.631
Combined Days	1.019	0.722	1.091	0.973	1.125	1.151	1.078	0.965
Combined Sexes								
Day 1	0.912	0.811	1.116	1.153	1.184	1.454	1.071	1.172
Day 4	1.035	0.569	1.008	0.692	1.104	0.864	1.049	0.719
Day 8	1.362	1.587	0.995	0.634	1.100	0.742	1.153	1.084
Combined Days	1.103	1.094	1.040	0.859	1.129	1.065		

TABLE 31

Means and standard deviations for mean spike amplitude (mv) for end of the first biceps brachii burst digital raw EMG signals across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	1.056	1.047	1.107	1.115	1.061	1.031	1.032	1.104	1.193	1.159
s.d.	0.790	0.878	1.018	1.173	1.109	0.985	0.801	0.994	1.233	1.051

TABLE 32

Analysis of variance for mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	1712.92756	56.27
SEX	1	.22050	.01
ERROR	14	30.44181	
DAYS	2	1.43745	.12
DS	2	3.23789	.27
ERROR	28	12.17441	
LOAD	2	1.01499	.21
LS	2	2.21243	.46
ERROR	28	4.84461	
DL	4	4.14789	1.07
DLS	4	1.63600	.42
ERROR	56	3.88644	
TRIAL	9	.43473	1.37
TS	9	.55716	1.76
ERROR	126	.31641	
DT	18	.24013	.85
DTS	18	.47586	1.69*
ERROR	252	.28135	
LT	18	.27776	1.36
LTS	18	.22495	1.10
ERROR	252	.20421	
DLT	36	.13497	.64
DLTS	36	.21503	1.02
ERROR	504	.21072	

* Significant at the .05 level

** Significant at the .01 level

the three test days of testing. There was no evidence to suggest that mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals changes with practice over days.

Main effect: trials. Mean spike aplitude for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and loads shown in Table 31, were greatest on trial nine (1.193 mv) and least on trial six (1.031 mv). With 9 and 126 degrees of freedom, an F of 1.95 is required for significance at the .05 level. The observed F of 1.37 failed to exceed this value necessary for significance. Thus, no significant differences existed between trials for mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals.

Main effect: sex. As seen in Table 30, mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals for combined test days, loads, and trials were similar for females (1.103 mv) and males (1.078 mv). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.01 did not exceed this required value for significance. Hence, mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals were not significantly different between females and males.

Main effect: loads. As shown in Table 30, mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and trials were

similar across load zero (1.103 mv), load one (1.040 mv) and load two (1.129 mv). With 2 and 28 degrees of freedom an F of 3.34 is required for significance at the .05 level. The observed F for between loads zero, one, and two of 0.21 failed to surpass this value required for significance. Therefore, there were no significant differences between loads zero, one, and two for mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals.

Interaction: days by trials by sex. With 18 and 252 degrees of freedom an F of 1.67 is required for significance at the .05 level. The observed F of 1.69 exceeded this value necessary for significance. Hence, mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals over trials for test days one, four, and eight did not follow the same patterns for each sex level. As shown in Figure 14, mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals remained relatively unchanged for females on test day one across trials compared to males who peaked dramatically on trials nine and ten. In addition, females appear to have reached two peaks for mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals on test days four and eight for trials four and nine as shown if Figure 14. No such tendency to attain distinct peaks of mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals on test days

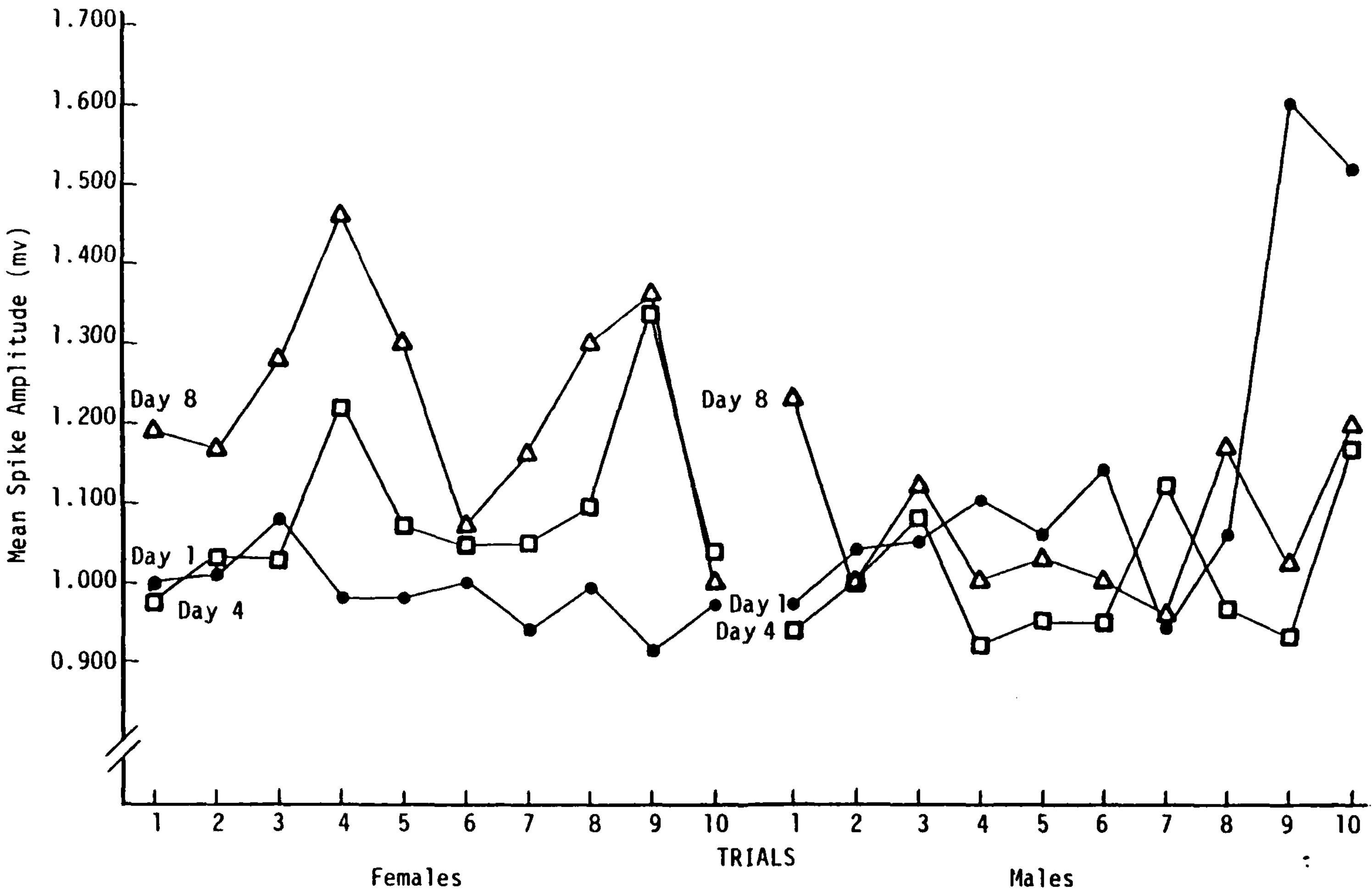

Figure 14. Mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals across trials for test days one, four, and eight for females and males.

four and eight was observed for males. As shown in Table 33, standard deviations for mean spike amplitude for end of the first biceps brachii burst digital raw EMG signals were generally high, as they often exceeded mean values.

Mean spike duration for end of the first biceps brachii burst digital raw EMG signals. Means and standard deviations of mean spike duration for end of the first biceps brachii burst digital raw EMG signals for test days one, four, and eight and combined days are presented in Table 34. Included in Table 34 are means and standard deviations of mean spike duration for end of the first biceps brachii burst digital raw EMG signals for females and males and combined sexes as well as loads zero, one, and two and combined loads for combined trials. Means and standard deviations for trials one through ten for combined test days, loads, and sexes are presented in Table 35. The four-factor analysis of variance of mean spike duration for end of the first biceps brachii burst digital raw EMG signals for day, sex, load, and trial main effects is presented is Table 36.

Main effect: test days 1, 4, 8. Mean spike duration for end of the first biceps brachii burst digital raw EMG signals for combined sexes, loads, and trials decreased from test day one (14.9 ms) to test day four (13.9 ms) to test day eight (13.2 ms). With 2 and 28 degrees of freedom an F of 5.49 is required for

TABLE 33

Means and standard deviations for mean spike amplitude (mv) for end of the first biceps brachii burst digital raw EMG signals across trials for test days one, four, and eight for females and males.

		Trials									
Females		1	2	3	4	5	6	7	8	9	10
Day 1	Mean	1.009	1.011	1.083	0.981	0.978	1.002	0.935	1.026	0.897	0.975
	s.d.	0.680	0.734	1.103	0.776	0.782	0.985	0.537	0.766	0.645	0.839
Day 4	Mean	0.977	1.036	1.048	1.220	1.062	1.054	1.055	1.090	1.336	1.057
	s.d.	0.728	0.815	0.702	1.087	0.842	0.880	0.925	0.831	1.185	0.721
Day 8	Mean	1.199	1.174	1.282	1.451	1.295	1.049	1.145	1.294	1.357	1.015
	s.d.	1.077	1.227	1.584	2.090	1.714	0.934	1.080	1.754	1.370	0.764
Males											
Day 1	Mean	0.971	1.040	1.049	1.112	1.059	1.137	0.941	1.071	1.611	1.528
	s.d.	0.870	1.231	1.183	1.286	1.608	1.705	0.957	1.062	2.202	1.963
Day 4	Mean	0.944	1.041	1.079	0.921	0.950	0.944	1.120	0.966	0.928	1.173
	s.d.	0.500	0.526	0.530	0.451	0.570	0.437	0.659	0.453	0.410	0.613
Day 8	Mean	1.234	1.008	1.101	1.008	1.027	1.004	0.945	1.180	1.030	1.206
	s.d.	0.795	0.516	0.710	0.585	0.634	0.546	0.538	0.621	0.561	0.790

TABLE 34

Means and standard deviations of mean spike duration (ms) for end of the first biceps brachii burst digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Loads	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	13.9	3.3	14.5	3.4	14.4	2.9	14.3	3.2
Day 4	14.6	3.5	14.5	3.5	13.2	2.8	14.1	3.3
Day 8	13.4	2.9	13.7	3.1	12.8	2.1	13.3	2.8
Combined Days	14.0	3.3	14.2	3.4	13.5	2.7	13.9	3.1
Males								
Day 1	16.4	4.9	15.8	6.0	14.4	3.6	15.5	5.0
Day 4	13.7	3.6	14.2	3.8	13.1	2.5	13.7	3.4
Day 8	13.2	3.4	13.1	3.5	13.2	2.6	13.2	3.2
Combined Days	14.4	4.2	14.4	4.7	13.5	3.0	14.1	4.1
Combined Sexes								
Day 1	15.2	4.3	15.2	4.9	14.4	3.3	14.9	4.2
Day 4	14.2	3.6	14.4	3.6	13.1	2.7	13.9	3.4
Day 8	13.3	3.1	13.4	3.3	13.0	3.4	13.2	3.0
Combined Days	14.2	3.8	14.3	4.1	13.5	2.9		

TABLE 35

Means and standard deviations for mean spike duration for *end of the first biceps brachii burst* digital raw EMG signals across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	14.5	14.1	14.5	13.9	13.9	13.7	14.0	14.0	14.0	13.7
s.d.	4.9	3.6	4.1	3.5	3.8	3.3	3.5	3.1	3.0	3.1

TABLE 36

Analysis of variance for mean spike duration for end of the first biceps brachii burst digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	282284.18853	829.21
SEX	1	19.00389	.06
ERROR	14	340.42703	
DAYS	2	345.03321	6.53**
DS	2	96.12029	1.82
ERROR	28	52.84834	
LOAD	2	90.73416	11.57**
LS	2	5.65816	.72
ERROR	28	7.83898	
DL	4	8.95547	.72
DLS	4	37.63960	3.01*
ERROR	56	12.49895	
TRIAL	9	10.32632	1.26
TS	9	6.56713	.80
ERROR	126	8.20985	
DT	18	7.82058	.96
DTS	18	8.54291	1.04
ERROR	252	8.18568	
LT	18	3.71864	.42
LTS	18	4.77293	.54
ERROR	252	8.84982	
DLT	36	8.00239	1.00
DLTS	36	5.70415	.71
ERROR	504	8.03209	

* Significant at the .05 level

** Significant at the .01 level

significance at the .01 level. The observed F of 6.53 exceeded this value necessary for significance. Hence, significant differences existed between test day levels for mean spike duration for end of the first biceps brachii burst digital raw EMG signals. Mean separation procedures revealed that test day four was not significantly different from either test day one or test day eight for mean spike duration for end of the first biceps brachii burst digital raw EMG signals. However, test days one and eight were significantly different for mean spike duration for end of the first biceps brachii burst digital raw EMG signals. Therefore, mean spike duration for end of the first biceps brachii burst digital raw EMG signals decreased 1.7 ms between test day one to test day eight. Standard deviations decreased over test days one (4.2), four (3.4), and eight (3.0).

Main effect: trials. Means for mean spike duration for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and loads presented in Table 35, were similar across all trials. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 1.26 failed to exceed this value necessary for significance. Hence, no significant differences existed between trials for mean spike duration for end of the first biceps brachii burst digital raw EMG signals.

Main effect: sex. As shown in Table 34, mean spike duration for end of the first biceps brachii burst digital raw EMG signals

for combined test days, loads, and trials were similar for females (13.9 ms) and males (14.1 ms). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.06 failed to surpass this value necessary for significance. Therefore, there were no significant differences between sexes for mean spike duration for end of the first biceps brachii burst digital raw EMG signals.

Main effect: loads. As seen in Table 34, mean spike duration for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and trials were lower for load two (13.5 ms) than loads zero (14.2 ms) and one (14.3 ms). With 2 and 28 degrees of freedom an F of 5.45 is required for significance at the .01 level. The observed F of 11.57 exceeded this value necessary for significance. Mean separation procedures revealed that loads zero and one were not significantly different from each other while load two was significantly different from both loads zero and one for mean spike duration for end of the first biceps brachii burst digital raw EMG signals. Thus, mean spike duration for end of the first biceps brachii burst digital raw EMG signals was significantly lower for load two than loads zero and one. Standard deviations were also lower for load two (2.9) than loads zero (3.8) and one (4.1).

Interaction: days by loads by sexes. With 4 and 56 degrees of freedom an F of 2.54 is required for significance at the .05

level. The observed F of 3.01 exceeded this value necessary for significance. Thus, mean spike duration for end of the first biceps brachii burst digital raw EMG signals over loads for test days one, four, and eight did not follow the same patterns for each sex level. As shown in Figure 15, mean spike duration for end of the first biceps brachii burst digital raw EMG signals increased from load zero to loads one and two on test day one for females while males decreased with increasing load on test day one. On test day four, similar decreases in mean spike duration for end of the first biceps brachii burst digital raw EMG signals were observed for males and females between loads one and two.

Mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals. Means and standard deviations of mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for test days one, four, and eight and combined days are presented in Table 37. Also in Table 37 are means and standard deviations of mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for males and females and combined sexes as well as loads zero, one, and two and combined loads and combined trials. Means and standard deviations for trial one through trial ten for combined test days, loads, and sexes are presented in Table 38. The four-factor analysis of variance of

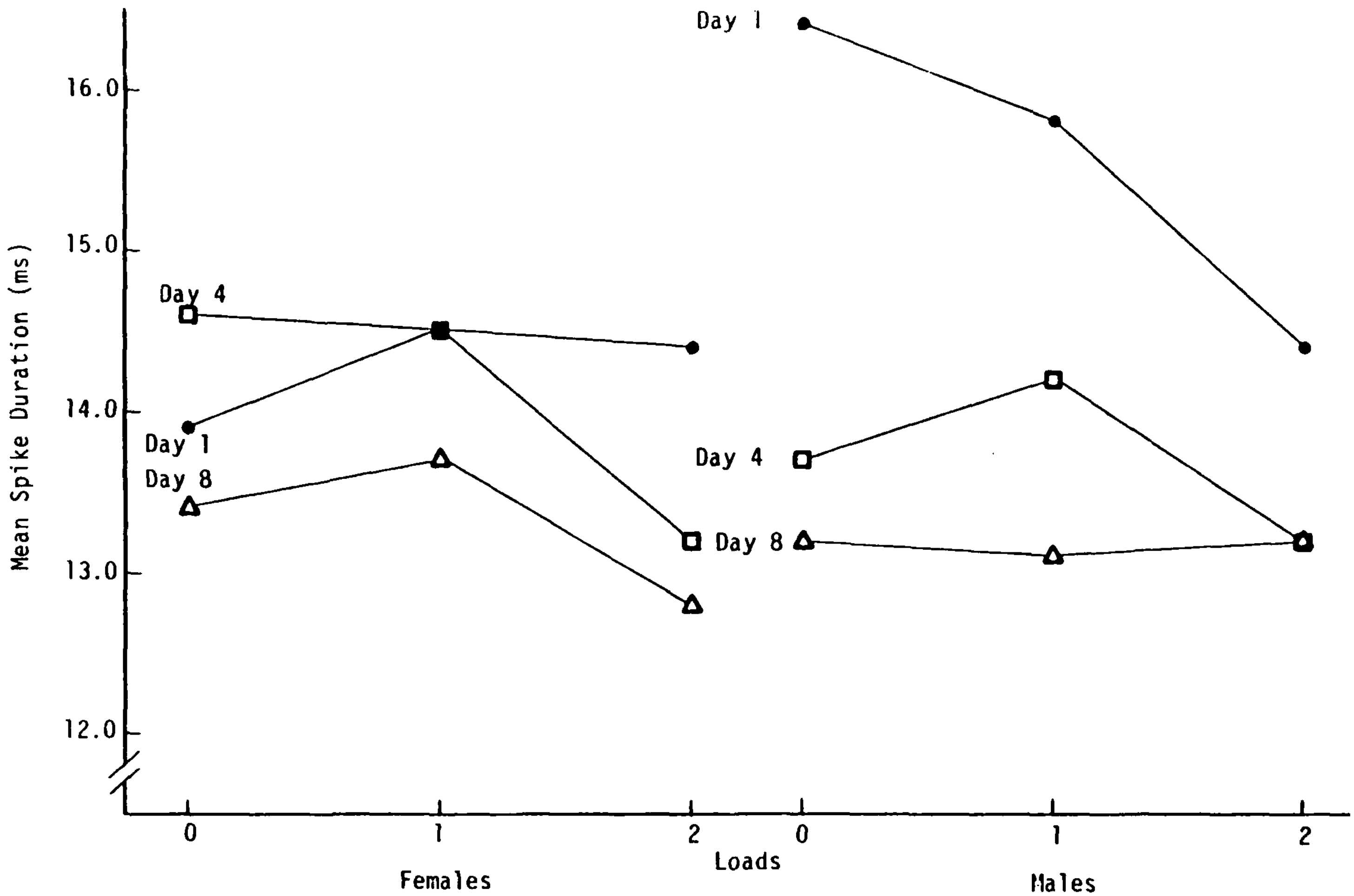

Figure 15. Mean spike duration for end of the first biceps brachii burst digital raw EMG signals across loads for test days one, four, and eight for females and males.

TABLE 37

Means and standard deviations of mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		*Load 1*		*Load 2*		*Combined Loads*	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	2.0	0.4	2.1	0.6	2.1	0.4	2.1	0.5
Day 4	2.1	0.6	2.1	0.5	2.0	0.5	2.1	0.5
Day 8	2.0	0.5	2.1	0.5	2.1	0.4	2.1	0.4
Combined Days	2.1	0.5	2.1	0.5	2.1	0.4	2.1	0.5
Males								
Day 1	2.1	0.7	2.1	0.6	2.0	0.5	2.0	0.6
Day 4	1.9	0.5	2.0	0.7	1.9	0.5	1.9	0.6
Day 8	1.9	0.7	1.8	0.5	1.9	0.5	1.9	0.6
Combined days	1.9	0.6	2.0	0.6	1.9	0.5	1.9	0.6
Combined Sexes								
Day 1	2.1	0.6	2.1	0.6	2.0	0.5	2.1	0.5
Day 4	2.0	0.5	2.1	0.6	2.0	0.5	2.0	0.5
Day 8	1.9	0.6	2.0	0.5	2.0	0.4	2.0	0.5
Combined Days	2.0	0.6	2.0	0.6	2.0	0.5		

TABLE 38

Means and standard deviations for mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	2.0	2.0	2.0	2.0	2.0	1.9	2.0	2.0	2.1	2.0
s.d.	0.6	0.6	0.5	0.5	0.5	0.5	0.5	0.5	0.6	0.5

TABLE 39

Analysis of variance for mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	5792.11544	2066.35
SEX	1	5.51236	1.97
ERROR	14	2.80306	
DAYS	2	1.03865	1.53
DS	2	1.02322	1.50
ERROR	28	.68003	
LOAD	2	.25811	1.22
LS	2	.01355	.06
ERROR	28	.21104	
DL	4	.25103	.81
DLS	4	.22685	.73
ERROR	56	.31137	
TRIAL	9	.20855	.83
TS	9	.19996	.80
ERROR	126	.25120	
DT	18	.35827	1.38
DTS	18	.40721	1.57
ERROR	252	.25906	
LT	18	.26911	1.09
LTS	18	.20166	.82
ERROR	252	.24692	
DLT	36	.27339	1.19
DLTS	36	.26691	1.16
ERROR	504	.23012	

* Significant at the .05 level

** Significant at the .05 level

mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for the main effects of days, sexes, loads, and trials is presented in Table 39.

Main effect: test days 1, 4, 8. Mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for combined sexes, loads, and trials were very similar over test days one (2.1 peaks), four (2.0 peaks) and eight (2.0 peaks). With 2 and 28 degrees of freedom, an F of 3.34 is required for significance at the .05 level. The observed F of 2.07 failed to exceed this value necessary for significance. Hence, there were no significant differences between mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals on the three test days of testing. No evidence was obtained suggesting that mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals changes with practice over days.

Main effect: trials. Mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for combined test days, loads, and sexes shown in Table 38 were very similar across all trials. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 0.83 failed to exceed this value necessary for significance. Thus, no significant differences existed for mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals between trials.

Main effect: sex. As shown in Table 37, mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for combined test days, loads, and trials were very similar for females (2.1 peaks) and males (1.9 peaks). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 1.97 did not exceed this value necessary for significance. No significant differences existed between males and females for mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals.

Main effect: loads. As seen in Table 37, mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and trials were the same across loads zero (2.0 peaks), one (2.0 peaks), and two (2.0 peaks). With 2 and 28 degrees of freedom an F of 3.34 is required at the .05 level. The observed F of 1.22 failed to surpass this value necessary for significance. Thus, mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals were not significantly different between loads.

Spike frequency for end of the first biceps brachii burst digital raw EMG signals. Means and standard deviations of spike frequency for end of the first biceps brachii burst digital raw

EMG signals for test days one, four, and eight and combined days are presented in Table 40. Included in Table 40 are means and standard deviations of spike frequency for end of the first biceps brachii burst digital raw EMG signals for females and males and combined sexes as well as loads zero, one, and two and combined loads and combined trials. Means and standard deviations for trial one through trial ten for combined test days, loads, and sexes are presented in Table 41. The four-factor analysis of variance of spike frequency for end of the first biceps brachii burst digital raw EMG signals for day, sex, load, and trial main effects are presented in Table 42.

Main effect: test days 1, 4, 8. Spike frequency for end of the first biceps brachii burst digital raw EMG signals for combined sexes, loads, and trials increased across test days one (71.6 Hz), four (76.0 Hz), and eight (79.2 Hz). With 2 and 28 degrees of freedom an F of 5.45 is required for significance at the .01 level. The observed F of 5.99 exceed this value necessary for significance. Thus, significant differences existed between test day levels for spike frequency for end of the first biceps brachii burst digital raw EMG signals. Mean separation procedures [51] revealed that test day four was not significantly different from test days one and eight while test day one was significantly different from test day eight for spike frequency for end of the first biceps brachii burst digital raw EMG signals. Spike frequency for end of the first biceps brachii

TABLE 41

Means and standard deviations for spike frequency for end of the first biceps brachii burst digital raw EMG signals across trials for combined test days, loads, and sexes.

	Trials									
Combined Days	1	2	3	4	5	6	7	8	9	10
Mean	74.2	75.3	73.9	75.9	77.1	77.1	75.9	75.1	75.2	76.2
s.d.	18.1	16.9	18.1	16.9	19.4	18.1	18.2	16.7	16.2	17.7

TABLE 40

Means and standard deviations of spike frequency for end of the first biceps brachii burst digital raw EMG signals under three loads and combined loads for females and males and combined sexes over test days one, four and eight and combined test days.

	Load 0		Load 1		Load 2		Combined Days	
Females	Mean	s.d.	Mean	s.d.	Mean	s.d.	Mean	s.d.
Day 1	75.6	16.4	72.6	17.3	72.2	14.1	73.5	16.0
Day 4	72.0	15.6	72.9	17.4	78.9	15.8	74.6	16.5
Day 8	78.1	16.0	76.5	15.6	80.5	14.0	78.4	15.2
Combined Days	75.2	16.1	74.0	16.8	77.2	15.0	75.5	16.0
Males								
Day 1	65.9	18.0	69.4	20.0	73.7	17.5	69.7	18.6
Day 4	78.1	21.2	74.8	17.6	79.5	15.3	77.5	18.2
Day 8	80.2	19.1	81.9	21.4	77.9	16.4	80.0	19.1
Combined Days	74.7	20.4	75.4	20.2	77.0	16.5	75.7	19.1
Combined Sexes								
Day 1	70.7	17.8	71.0	18.6	73.0	15.8	71.6	17.4
Day 4	75.0	18.8	73.8	17.5	79.2	15.5	76.0	17.4
Day 8	79.2	17.5	79.2	18.9	79.2	15.3	79.2	17.3
Combined Days	75.0	18.4	74.7	18.6	77.1	15.8		

TABLE 42

Analysis of variance for spike frequency for end of the first biceps brachii burst digital raw EMG signals

SOURCE	DEGREES OF FREEDOM	MEAN SQUARE	F
MEAN	1	8227843.66248	972.14
SEX	1	20.01315	.00
ERROR	14	8463.59728	
DAYS	2	7058.05357	5.99**
DS	2	1525.82662	1.29
ERROR	28	1178.81796	
LOAD	2	853.39994	3.83*
LS	2	126.30145	.57
ERROR	28	223.09334	
DL	4	323.37697	.90
DLS	4	1073.01807	2.98*
ERROR	56	360.20966	
TRIAL	9	163.91395	.90
TS	9	120.72358	.66
ERROR	126	182.15337	
DT	18	173.70136	.86
DTS	18	217.26177	1.07
ERROR	252	202.99137	
LT	18	147.58885	.70
LTS	18	90.85794	.43
ERROR	252	211.96554	
DLT	36	187.26305	1.04
DLTS	36	180.13726	1.00
ERROR	504	179.55709	

* Significant at the .05 level

** Significant at the .01 level

burst digital raw EMG signals significantly increased be test day one to test day eight. Standard deviations were similar on test days one (17.4), four (17.4), and eight (17.3).

Main effect: trials. Means for spike frequency for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and loads presented in Table 41, were similar across all trials. With 9 and 126 degrees of freedom an F of 1.95 is required for significance at the .05 level. The observed F of 0.90 failed to exceed this value necessary for significance. Hence, no significant differences existed between trials for spike frequency for end of the first biceps brachii burst digital raw EMG signals.

Main effect: sex. As shown in Table 40, spike frequency for end of the first biceps brachii burst digital raw EMG signals for combined test days, loads, and trials were similar for females (75.5 Hz) and males (75.7 Hz). With 1 and 14 degrees of freedom, an F of 4.60 is required for significance at the .05 level. The observed F of 0.00 failed to exceed this value necessary for significance. Thus, no significant differences existed between sexes for spike frequency for end of the first biceps brachii burst digital raw EMG signals.

Main effect: loads. As seen in Table 40, spike frequency for end of the first biceps brachii burst digital raw EMG signals for combined test days, sexes, and trials were similar for loads zero (75.0 Hz) and one (74.7 Hz) and greatest for load two (77.1 Hz). With 2 and 28 degrees of freedom an F of 3.34 is required

for significance at the .05 level. The observed F of exceeded this value necessary for significance. Mean separation procedures [51] revealed that loads zero and one were not significantly different from each other while, loads zero and one were significantly different from load two for spike frequency for end of the first biceps brachii burst digital raw EMG signals. Spike frequency for end of the first biceps brachii burst digital raw EMG signals was greater under load two than loads zero and one. Standard deviations were slightly lower under load two (15.8) than loads zero (18.4) and one (18.6).

Interaction: days by loads by sexes. With 4 and 56 degrees of freedom an F of 2.54 is required for significance at the .05 level. The observed F of 2.98 exceeded this value necessary for significance. Hence, spike frequency for end of the first biceps brachii burst digital raw EMG signals over loads for test days one, four, and eight did not follow the same patterns for each sex level. As shown in Figure 16, females tended to decrease spike frequency from load zero to loads one and two on test day one while, males increased in spike frequency across loads zero to two. On test days four and eight, females generally increased in spike frequency from load zero to load two while, males displayed no discernible pattern.

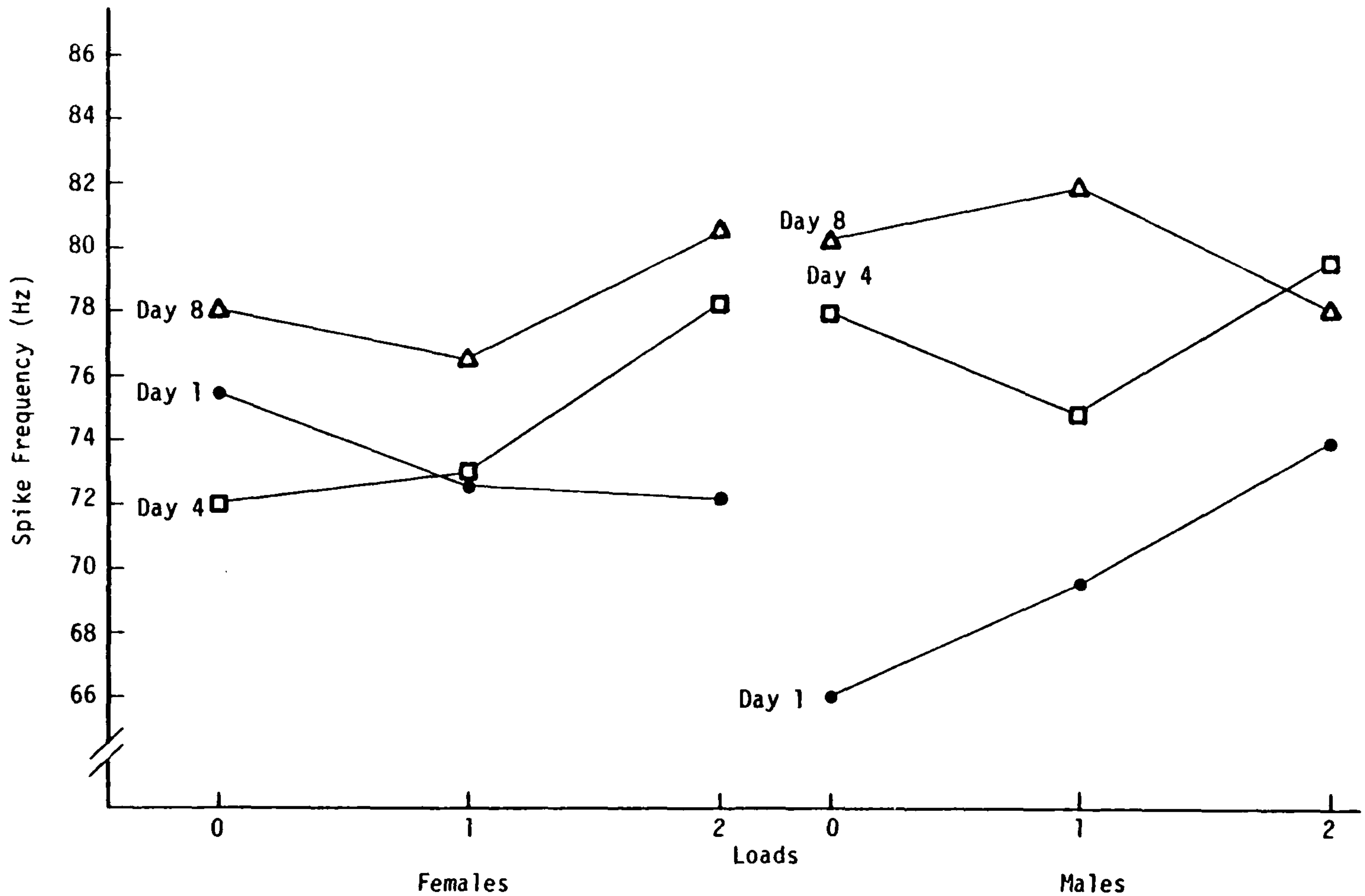

Figure 16. Spike frequency for end of the first biceps brachii burst digital raw EMG signals across loads for test days one, four and eight for females and males.

Intercorrelations Between Maximum Speed Forearm Flexion Movement time and Percent Acceleration Time, Number of Spikes, Mean Spike Amplitude, Mean Spike Duration, Mean Number of Peaks Per Spike, and Spike Frequency for Biceps Brachii Motor Time and End of the First Biceps Burst Digital Raw EMG Signals for Test Day Eight

Pearson correlation coefficients between maximum speed forearm flexion movement time and percent acceleration time, number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike, and spike frequency for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals are presented in Table 43. The data to be analyzed were recorded from eight male and eight female subjects on the last day of testing (test day 8). With 14 degrees of freedom, r's of .497 and .623 are required for significance at the .05 and .01 level respectively.

Among EMG spike parameters, significant positive correlations were observed between biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals for number of spikes (r=.52), mean spike amplitude (r=.86), mean spike duration (r=.74), and spike frequency (r=.70). Number of

TABLE 43

Intercorrelations of maximum speed forearm flexion movement time and percent acceleration time, number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike and spike frequency for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals for test day eight.

	2	3	4	5	6	7	8	9	10	11	12
Movement Time (1)	-.56*	.28	.57*	.19	.29	.20	.17	-.19	.28	-.22	-.20
% Acceleration Time (2)		-.43	-.65*	-.11	-.32	.25	.37	.19	-.12	-.23	-.32
Number of Spikes - Biceps Motor Time (3)			.52*	.23	.30	-.72*	-.44	-.36	.26	.72*	.43
Number of Spikes - End of the First Biceps Burst (4)				-.24	.01	-.43	-.53*	.03	.44	.40	.46
Mean Spike Amplitude - Biceps Motor Time (5)					.86*	.31	.35	-.47	-.36	-.31	-.31
Mean Spike Amplitude - End of the First Biceps Burst (6)						.14	.22	-.23	-.11	-.15	-.21
Mean Spike Duration - Biceps Motor Time (7)							.74*	.13	-.28	-.99*	-.71*
Mean Spike Duration - End of the First Biceps Burst (8)								.02	.01	-.72*	-.98*
Mean Number of Peaks Per Spike - Biceps Motor Time (9)									.42	-.14	.01
Mean Number of Peaks Per Spike - End of the First Biceps Burst (10)										.27	-.04
Spike Frequency - Biceps Motor Time (11)											.70*
Spike Frequency - End of the First Biceps Burst (12)											

*N = 16. Pearson r required for significance: $r_{.01} = .623$; $r_{.05} = .497$

spikes for biceps brachii motor time digital raw EMG signals demonstrated a significant negative correlation with mean spike duration for biceps brachii motor time (r=-.72) digital raw EMG signals as well as a positive correlation with spike frequency for biceps brachii motor time (r=.72) digital raw EMG signals. Number of spikes for end of the first biceps brachii burst digital raw EMG signals demonstrated a negative correlation with end of the first biceps brachii burst (r=-.53) digital raw EMG signals. As would be expected, mean spike duration and spike frequency were highly negatively correlated when recorded from the same digital raw EMG signal (biceps brachii motor time, r=-.99; end of the first biceps brachii burst, r=-.98). Lower correlation coefficients were observed between mean spike duration for biceps brachii motor time digital raw EMG signals and spike frequency for end of the first biceps brachii burst digital raw EMG signals (r=-.71) as well as between mean spike duration for end of the first biceps brachii burst and spike frequency for biceps brachii motor time digital raw EMG signals (r=-.72). In general, higher correlation coefficients were observed between EMG spike parameters recorded from the same digital raw EMG signal rather than different digital raw EMG signals.

Of particular importance to this study is that maximum speed forearm flexion movement time demonstrated a significant negative correlation (r=-.56) with maximum speed forearm flexion percent

acceleration time and a significant positive correlation (r with number of spikes for end of the first biceps brachii burst digital raw EMG signals. Likewise, maximum speed forearm flexion percent acceleration time negatively correlated with number of spikes for end of the first biceps brachii burst (r=-.65) digital raw EMG signals. No other EMG spike parameters significantly correlated with maximum speed forearm flexion movement time or maximum speed foream flexion percent acceleration time.

DISCUSSION

Reliability

Maximum speed forearm flexion movement time. The intraclass correlation coefficient for maximum speed forearm flexion movement time under load zero (r=.97) was similar to the coefficient reported by Wolcott [95] of .96 and greater than the coefficient reported by Lagasse [47] of .88 for maximum speed forearm flexion. In the present study, examination of variance components shows a much greater true score variance (S^2_t=582.57) than that found by Lagasse (S^2_t=214.00) or Wolcott (S^2_t=174.68). A possible explanation for these observed differences in true score variance may be found in that both male and female subjects were used in the present study.

In Wolcott's [95] study, lower true score variance was offset by a substantially lower trial error variance (S^2_t=16.72) relative to the present study (S^2_t=65.07) resulting in similar intraclass reliability coefficients in both studies under unloaded conditions. This difference in trial error variance may be due to the different movement tasks utilized in both studies.

While both studies involved maximum speed forearm flexions, Wolcott's subjects were not required to culminate their movement at a specific target. Requiring culmination of elbow flexion at a target adds complexity to the movement and therefore may have made movement replication more difficult in the present investigation.

In the present study, an overall increase in trial error variance (S^2_t=65.07, S^2_t=55.29, S^2_t=96.23) and day error variance (S^2_d=28.28, S^2_d=97.93, S^2_d=107.69) with a concurrent increase in true score variance (S^2_t=582.57, S^2_t=673.94, S^2_t=810.94) for maximum speed forearm flexion movement time resulted in similarly high intraclass correlation coefficients (r=.97, r=.93, r=.93) for loads zero, one, and two respectively. As previously reported by Wolcott [95], these observed increases in variance components with increasing load indicate that greater variability was associated with moving heavier loads. Moving heavier loads seems to have resulted in greater inter-subject variability as evidenced by the observed increases in true score variance with increasing load.

Maximum speed forearm flexion percent acceleration time. The intraclass reliability coefficient observed for maximum speed forearm flexion percent acceleration time for load zero (r=.70) was lower than has been previously reported by Lagasse [47] of .81 and Wolcott [95] of .85. In the present study, intraclass

reliability coefficients were observed to increase from load one (r=.64) to load two (r=.86). In contrast, Wolcott [95] reported nearly equivalent intraclass reliability coefficients for maximum speed forearm flexion percent acceleration time under increasing load conditions (r=.79; r=.78). The difference between the intraclass reliability coefficients observed in the present study and those reported by Wolcott [95], may be partially due to the previously mentioned differences in movement task performed.

Inspection of variance components for maximum speed forearm flexion percent acceleration time shown in Table 3 revealed that more error variance was trial error variance rather than day error variance. This finding is in agreement with findings reported by both Wolcott [95] and Lagasse [47] and suggests that future studies measuring this parameter should focus on increasing the number of trials used as a means of improving reliability.

EMG spike parameters. Fundamental in EMG research is the question of reproducibility of measurements. A few studies have reported data emphasizing the technical care required to obtain reliable EMG recordings [44,64,84]. Using bipolar surface electrodes similar to those used in the present study, Komi and Buskirk [44] and Pancherz and Winnberg [64] have demonstrated that reliable integrated EMG measurements can be obtained across days for maximal concentric contractions. However, the

applicability of these findings to the more detailed EMG analysis employed in the present study is limited. In a later study, Viitasalo and Komi [84] reported findings of the reproducibility of measurements of select EMG signal characteristics taken during submaximal and maximal isometric contractions. They were able to show that EMG spike parameters such as number of spikes, spike amplitude and spike slope could be reliably measured between test sessions within the same day. However, spike slope and spike amplitude reliability coefficients determined for day to day measurements were relatively poor.

The present investigation represents a first attempt to reliably measure selected EMG spike parameters recorded during ballistic muscular contractions. Consequently, the results of the reliability analysis performed for the EMG spike parameters measured in this study is of special significance.

Interestingly, every EMG spike parameter was reliable for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals with the exception of mean spike slope which was never reliable. No one of the EMG spike parameters was reliable for second biceps brachii burst and triceps brachii burst digital raw EMG signals. A possible explanation for the higher reliability of EMG spike parameters for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals may lie in the fact that these signals were sampled from larger biceps brachii digital raw EMG

signals by visually selecting their duration relative to movement initiation event markers (See Figure 7,page 62). Digital raw EMG signals sampled relative to a movement event marker could have a lower probability of sampling error compared to samples taken of an entire EMG burst as was done for the second biceps brachii burst and the triceps brachii burst. Future studies wishing to improve the reliability of EMG data take from the second biceps brachii burst and triceps brachii burst should take samples relative to the movement completion event marker. Sampling EMG signals this way, while increasing the number of trials performed, would help to lower the generally high trial error variance observed of many of the EMG spike parameters used in this study.

Influence of Practice Over days on Kinematic Parameters

Maximum speed forearm flexion movement time. The overall 15 ms (10%) decrease in maximum speed forearm flexion movement time observed over practice days one through eight (80 trials) is in general agreement with numerous investigations [12,23,42,43,47,48,62,66,95] and reconfirms the well known belief that practice improves performance. Lagasse [47] reported a significant (5%) decrease in movement time over two practice days

(100 trials) of 7 ms. Wolcott [95] observed a 13 ms (9%) decrease in movement time over four practice days (60 trials). While Lagasse [47] and Wolcott [95] used male subjects, Teves [81] used female subjects and found that 50 trials of maximum speed of forearm flexion practice over two practice days had no significant effect on movement time. Teves [81] suggests that female subjects may improve movement speed more rapidly (trial to trial rather than day to day) than male subjects. This issue will be addressed in a later section in this discussion on the influence of gender on kinematic parameters.

Significant decreases in maximum speed forearm flexion movement time were observed, in the present study, between test days four and eight in addition to between test days one and four. The 7.7 ms (5%) drop in maximum speed forearm flexion movement time between test days four and eight was approximately the same as the 6.9 ms (5%)drop observed between test days one and four. This finding is not in accord with data reported by Wolcott [95] in which significant decreases in movement time were observed exclusively over the first four practice days after which no further decreases in movement time were observed. A possible explanation for the decreases in maximum speed forearm flexion movement time observed in the present study over test days four and eight may be found in the differences between the movements studies in both investigations. The class B (antagonist arrested) maximum speed forearm flexion movement

utilized in the present study probably requires more complex alterations in central motor commands during practice than the less complex class A (artificially arrested) movement used by Wolcott [95]. The more complex class B movement may have a greater capacity for improvement and may therefore improve over a longer period of time.

Maximum speed forearm flexion percent acceleration time. Although an increase of 5 percent acceleration time was observed over practice days one through eight, the increase was not statistically significant. This result is not in accord with findings previously reported by Wolcott [95] and Lagasse [47], using male subjects, who found increases of 17 percent and 14 percent acceleration time over four practice days (60 trials) and two practice days (100 trials) respectively. However, no significant changes in maximum speed forearm flexion percent acceleration time over practice days observed in the present study is in agreement with Teve's [81] study in which female subjects were tested after performing 50 trials of maximum speed forearm flexion over two days. Considering the results reported by Teves [81], it seems reasonable to suggest that the lack of significant change in maximum speed forearm flexion percent acceleration time may be partially explained by the fact that eight of the sixteen subjects used in the present study were females.

Of major significance to the present study are the implications of observing no change in maximum speed forearm flexion percent acceleration time concurently with decreases in maximum speed forearm flexion movement time. This combination can only occur if the magnitude of acceleration of the forearm increased with practice across test days. In contrast, increases in maximum speed forearm flexion percent acceleration time with simultaneous decreases in maximum speed forearm flexion movement time, as reported by Wolcott [95], can only occur if the period of time over which deceleration of the forearm occurs decreased with practice over test days. Changes in acceleration were not assessed in Wolcott's [95] and Lagasse's [47] studies since their observed increases in maximum speed forearm flexion percent acceleration time may or may not have occured with concurrent increases in acceleration of the forearm. Increases in acceleration of the forearm with practice over days observed in the present study, are in agreement with data reported by Boucher and Lagasse [13] in which increases in forearm acceleration were observed with practice over days of a horizontal maximum speed forearm flexion task. Such increases in forearm acceleration (actually angular acceleration) over days with practice are highly suggestive of muscle tension increases (expressed as forearm torque) produced by the biceps brachii muscle during the isometric phase (biceps brachii motor time) and or during the isotonic phase (end of the first biceps brachii burst) since

forearm torque produced by the biceps brachii equals the forearm moment of inertia (constant over days) times forearm angular acceleration. The role of motor unit activation patterning during alterations in tension development capabilities of the biceps brachii muscle during practice over days will be examined in the next section of this discussion dealing with the influence of practice over days on EMG spike parameters.

Influence of Practice Over Days on EMG Spike Parameters

Introduction. It is known from temporal EMG studies of various maximal speed forearm flexion tasks that biceps brachii motor time does not significantly change with practice over days [12,47,95]. It has also been demonstrated that the total EMG activity time of the biceps brachii decreases with practice over days [12]. Since the total EMG activity time of the biceps brachii muscle is a composite of the biceps brachii motor time plus the end of the first biceps brachii burst, Boucher [12] has suggested that the duration of the end of the first biceps brachii burst decreases with practice over days.

Attempts to investigate gross quantitative EMG alterations during practice of maximum speed tasks have indicated that the magnitude of the total electrical activity (integrated EMG) of

the agonist muscle increases with practice over days [46]. Changes in EMG spike parameters during practice of a maximum speed task have been assessed for the first time in the present study.

Biceps brachii motor time. Both mean spike duration and spike frequency were observed to significantly change over test days for biceps brachii motor time digital raw EMG signals. Mean spike duration was observed to decrease by 1.3 ms (12%) between test days one and four and stabilized between test days four and eight. Conversely, spike frequency increased by 7.3 Hz (8%) between test days one and four and remained stable between test days four and eight.

No significant changes over test days were observed for number of spikes, mean spike amplitude, and mean number of peaks per spike. However, the pattern across test days for number of spikes was different for males than for females. Females demonstrated a decrease of 0.9 spikes (12%) while males demonstrated an increase of 0.6 spikes (8%) over test days one and four.

These data demonstrate for the first time that observable alterations occur in biceps brachii motor time raw EMG signal characteristics (as assessed by specific EMG spike parameters) during practice of a maximal speed forearm flexion task over days. In addition, these data suggest that increases in total

electrical activity observed by Kots [46] over practice days are more the result of increases in spike frequency than spike amplitude.

The role of motor unit activation patterning during biceps brachii motor time in improving maximal speed forearm flexion movement time during practice over days can also be postulated from these results. The observed increases in spike frequency with concurrent decreases in mean spike duration and no change in mean spike amplitude suggests that firing rate modulation may be the most prominent mechanism involved in alterations in motor unit activation patterning for biceps brachii motor time during daily practice of a maximal speed forearm flexion task. Lending some support to this hypothesis, Komi and Viitasalo [45] have demonstrated that the number of spikes recorded per constant unit of time (i.e. spike frequency) increases as isometric muscle tension increases. Since an increase in muscle tension (measured as limb torque) must have occurred in the present study across days (because increases in forearm acceleration were observed), concurrent increases in spike frequency would be compatible with their findings.

Enhanced recruitment of higher threshold motor units with practice across days is also a possible mechanism since higher threshold motor units are correlated with higher maximum firing frequencies [35]. However, this possibility is less likely since recruitment of higher threshold motor units is also correlated

with higher spike amplitudes [57]. A trend toward motor unit synchronization is even less likely to have occurred with practice across days since motor unit synchronization is characterized by decreases in spike frequency [70] and mean number of peaks per spike [54] as well as increases in mean spike duration [54] and mean spike amplitude [15,50,52,58]. Based on these results, it is most likely that during biceps brachii motor time the same motor unit pool is being recruited at higher firing frequencies from day to day during practice.

<u>End of the first biceps brachii burst</u>. Mean spike duration and spike frequency were observed to significantly change across test days for end of the first biceps brachii burst digital raw EMG signals. In distinct contrast with biceps brachii motor time, however, these changes were statistically significant between test days one and eight and <u>not</u> statistically significant between test days one and four. The 1 ms (7%) decrease in mean spike duration between test days one and four was not statistically significant whereas, the 1.7 ms (13%) decrease in mean spike duration between test days one and eight was statistically significant. Similarly, the observed increases in spike frequency of 4.4 Hz (6%) between test days one and four was not statistically significant while the observed 7.6 Hz (11%) increase between test days one and eight was statistically significant. No significant changes over test days were observed

for number of spikes, mean spike amplitude, and mean number of peaks per spike.

These results demonstrate for the first time that daily practice of a maximal speed forearm flexion task elicits observable alterations in raw EMG signal characteristics (as assessed with specific EMG spike parameters) during the end of the first biceps brachii burst. It is interesting to note that significant increases in spike frequency and decreases in mean spike duration required more practice days to occur for end of the first biceps brachii burst digital raw EMG signals than biceps brachii motor time digital raw EMG signals. This finding suggests the possibility that daily practice effects on central motor commands to the biceps brachii muscle may be primarily in terms of mean spike duration decreases, and spike frequency increases for biceps brachii motor time and primarily in terms of total activity time decreases for the end of the first biceps brachii burst with mean spike duration decreases and spike frequency increases being of secondary significance.

Again, as for biceps brachii motor time, the end of the first biceps brachii burst was characterized by increases in spike frequency and decreases in mean spike duration even though these changes developed more slowly. This finding suggests that firing rate modulation may be a prominent mechanism of motor unit activation patterning during the end of the first biceps brachii burst but this mechanism may be less prominent than alterations

in the duration of the end of the first biceps brachii burst.

Influence of Practice Over Trials on Kinematic Parameters

The observed 11 ms (7%) decrease in maximum speed forearm flexion movement time over trials one through ten is in general agreement with Lagasse [47] who has reported a significant linear decrease in maximum speed forearm flexion movement time over trials. Also notable in the present study is the finding of a day-trial interaction that suggests that short term (trial to trial) decreases in maximum speed forearm flexion movement time occur to a greater extent over the first few trials on test day one than on subsequent test days (see Figure 10). This finding reconfirms Lagasse's [47] observation of significant decreases in maximum speed forearm flexion movement time over the first ten trials of practice on the first day of practice.

The stability of maximum speed forearm flexion percent acceleration time over trials with concurrent decreases in maximum speed forearm flexion movement time observed in the present study is analogous to the situation previously examined in this discussion on the influence of practice over days on kinematic parameters. This combination can occur only if acceleration of the forearm increased with practice across

trials. Therefore, the magnitude of the torque produced by the biceps brachii muscle (and other agonists) must have increased across trials one to ten.

Influence of Practice Over Trials on EMG Spike Parameters

No significant alterations in EMG spike parameters were observed across trials for both biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals with the exception of mean spike amplitude for biceps brachii motor time digital raw EMG signals. However, despite observing statistically significant differences for mean spike amplitude between trials, mean separation procedures revealed no discernible pattern of change of mean spike amplitude across trials (see Table 15). No reasonable explanation can be forwarded to explain this occurrence.

The fact that EMG spike parameters were not observed to change over trials suggests that alterations in central motor commands other than the EMG spike parameters studied in this investigation are most likely instrumental in the observed decrease in maximal speed forearm flexion movement time, no change in maximal speed forearm flexion percent acceleration time and therefore increase in magnitude of acceleration of the

forearm over trials. Other feasible mechanisms include decreases in biceps brachii motor time over trials reported by Lagasse [47] which was attributed to warm-up effects. It is possible that this decrease in biceps brachii motor time may be the mechanism most active during practice over trials.

Influence of Gender on Kinematic Parameters

Maximum speed forearm flexion movement times observed for females (176.7 ms) were 40.6 ms (23%) longer than for males (136.3 ms). Thus, male subjects were in general faster than female subjects for combined test days, loads and trials. Previous investigations have studied either males [47,95] or females [81] with the exclusion of the opposite sex and have not made direct comparisons between the sexes.

Maximum speed forearm flexion percent acceleration times observed for females (77.3%) were 10.4% less than for males (87.7%). In general, therefore, males accelerated their forearms for a larger percentage of maximum speed forearm flexion movement time than did females. This finding is compatible with the lower overall maximum speed forearm flexion movement times observed in males relative to females since, holding all else constant, a limb that accelerates longer will attain a higher maximum angular

velocity and therefore take less time to complete the movement.

Influence of Gender on EMG Spike Parameters

None of the EMG spike parameters were significantly different between the sexes for biceps brachii motor time digital raw EMG signals; however, number of spikes for end of the first biceps brachii burst digital raw EMG signals was 36% greater for females (8.8 spikes) than males (5.6 spikes). Since mean spike duration was not different between males and females, this finding suggests that females activate motor units over a longer time than males during the end of the first biceps brachii burst (dynamic phase).

The number of spikes for biceps brachii motor time digital raw EMG siginals generally increased with increasing load for both males and females. However, as can be seen in Figure 8, males increased to a greater extent under load two than did females suggesting an equalizing effect on the number of spikes for biceps brachii motor time. It appears that males require a lower number of spikes during biceps brachii motor time under lighter loads zero and one than females but require about the same number of spikes as females under heavier loads (load two). This finding suggests that since mean spike duration was not

different between the sexes for biceps brachii motor time digital raw EMG signals, males activate motor units for a shorter period of time than females at lighter inertial loads. These results are compatible with data reported by Visser and DeRijke [85] demonstrating that as a rule females require higher number of spikes than males to produce a given isometric tension.

Influence of Inertial Loads on Kinematic Parameters

Maximal speed forearm flexion movement time was 9.7 ms (7%) longer for load one (150.0 ms) than load zero (140.3 ms) and 29.5 ms (16%) longer for load two (179.5 ms) than for load one (150.0 ms). This finding was expected since forearm angular acceleration equals forearm torque divided by the forearm moment of inertia. Holding forearm torque constant, an increase in forearm moment of inertia should decrease the angular acceleration of the forearm thereby resulting in a decrease in maximum speed forearm flexion movement time at higher inertial loads. Inspection of Wolcott's [95] means for maximum speed forearm flexion movement time suggests a similar trend of increasing maximum speed forearm flexion movement time with increasing inertial load even though Wolcott did not test for a load main effect.

Maximum speed forearm flexion percent acceleration time was similar for lighter loads zero (84.1%) and one (84.7%) but less for heavier load two (78.7%). This finding suggests that imposing relatively heavier loads on a maximum speed forearm flexion task has the effect of decreasing the percentage of the total movement time during which acceleration of the forearm occurs. Inspection of means reported by Wolcott [95] for percent acceleration time measured under unloaded (89.5%) and lightly loaded (88.5%) conditions relative to heavily loaded (82.2%) conditions suggests similar decreases in maximum speed forearm flexion percent acceleration time for heavier loads even though no tests were performed to assess a load main effect.

Influence of Inertial Loads on EMG Spike Parameters

The number of spikes for biceps brachii motor time digital raw EMG signals were observed to be about fifteen percent greater for load two (8.5 spikes) than loads zero (7.2 spikes) and one (7.7 spikes). Inertial load had no significant effect on spike frequency, mean spike duration, mean number of peaks per spike, and mean spike amplitude for biceps brachii motor time digital raw EMG signals. The greater number of spikes and stable mean spike duration and spike frequency observed for biceps brachii

motor time digital raw EMG signals suggests that biceps brachii motor time was longer for the higher inertial load. This hypothesis is compatible with means for biceps brachii motor time reported by Wolcott [95] for maximum speed forearm flexion under increasing inertial load (69.29 ms, 76.75 ms, 80.58 ms). These data accentuate the interdependence between temporal and quantitative EMG measures.

In a similar fashion as for biceps brachii motor time digital raw EMG signals, the effect of inertial load on EMG spike parameters for end of the first biceps brachii burst digital raw EMG signals included an approximately (17%) greater number of spikes for load two (8.1 spikes) than for loads zero (6.7 spikes) and one (6.9 spikes). Unlike biceps brachii motor time digital raw EMG signals, spike frequency was slightly (about 3%) greater under load two (77.1 Hz) than loads zero (75.0 Hz) and one (74.7 Hz) while, mean spike duration was slightly shorter (about 5%) under load two (13.5 ms) than loads zero (14.2 ms) and one (14.3 ms). Mean spike amplitude and mean number of peaks per spike were not significantly altered by imposed inertial loads. Interpretation of these results is difficult since mean spike duration decreased when number of spikes increased with increasing inertial load.

Interrelationships Between Kinematic and EMG Spike Parameters

As was expected, maximum speed forearm flexion percent acceleration time significantly correlated with maximum speed forearm flexion movement time ($r=-.56$, $p\leq.05$). This negative correlation coefficient between maximum speed forearm flexion movement time and maximum speed forearm flexion percent acceleration time is greater than correlation coefficients previously reported by Wolcott [95] for a maximum speed forearm flexion task between these two parameters ($r=-.46$, $r=-.42$, $r=-.49$, $r=-.37$, $r=-.38$, $r=-.41$). A negative correlation between maximum speed forearm flexion movement time and maximum speed forearm flexion percent acceleration time is logical since a forearm that accelerates for a longer percentage of the movement time will be completed in a shorter period of time if all other variables are held constant.

In general, each EMG spike parameter positively correlated with itself between biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals. This finding suggests biceps brachii motor time and end of the first biceps brachii burst tend to be similar in terms of the EMG spike parameters measured in this study.

The observation that maximum speed forearm flexion movement time positively correlated with number of spikes for end of the

first biceps brachii burst digital raw EMG signals ($r=.57$, $p\leq.05$) but did not correlate with any of the other EMG spike parameters suggests that maximum speed forearm flexion movement time is not largely influenced by EMG spike parameters recorded from biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals. Such results are paradoxical since they imply that other neuromotor coordination mechanisms such as agonist-antagonist latency which are based on temporal EMG analysis techniques offer greater predictive power of maximum speed of human movement than quantitative EMG analyses which offer substantially more detailed information about motor unit firing patterns. The possibility exists that maximum speed forearm flexion movement time may more closely correlate with EMG spike parameters for the triceps brachii burst. It is unfortunate that it was not possible to test this hypothesis since EMG spike parameters measured during the triceps brachii burst were unreliable.

Maximum speed forearm flexion movement time may be indirectly influenced by maximum speed forearm flexion percent acceleration time which was found to be negatively correlated with number of spikes for end of the first biceps brachii burst ($r=-.65$) digital raw EMG signals. Therefore, a lower number of spikes recorded during the end of the first biceps brachii burst was associated with a higher percentage of maximum speed forearm flexion movement time during which acceleration of the forearm occurred.

SUMMARY

In the past, temporal electromyographic techniques have been applied to the study of the maximum speed of human movement phenomenon and have offered useful but limited insights into the detailed alterations in central motor commands during gross motor learning. This study was a first attempt at applying an unique assemblage of quantitative electromyographic techniques to the study of maximum speed of human movement. The primary purpose of this study was to assess the long term (day to day) alterations of two kinematic parameters and six EMG spike parameters during practice of a maximum speed forearm flexion task. The six EMG spike parameters studied were measured on four temporal EMG components (digital raw EMG signals) of the triphasic EMG pattern characteristic of ballistic flexions of the forearm. The four temporal EMG components of the triphasic EMG pattern studied in this investigation included: biceps brachii motor time; end of the first biceps brachii burst; second biceps brachii burst; and triceps brachii burst digital raw EMG signals. Secondary problems investigated in the present study included: the influence of short term (trial to trial) practice on kinematic and EMG spike parameters; the influence of gender on kinematic

and EMG spike parameters; and the influence of three variations of inertial loading on kinematic and EMG spike parameters. The two kinematic parameters measured were maximum speed forearm flexion movement time and maximum speed forearm flexion percent acceleration time. The six EMG spike parameters measured were number of spikes, mean spike amplitude, mean spike duration, mean spike slope, mean number of peaks per spike, and spike frequency.

Sixteen college age students, eight male and eight female participated in this study. Each subject performed ten maximum speed forearm flexion trials under each of three different inertial loads on eight separate practice days. At least 30 seconds separated trials to insure maintainence of a non-fatigued state. The first three practice days were separated by at most 24 hours while the last five practice days were separated by at most 48 hours. Data were recorded on three of the eight practice days (i.e., test days one, four, and eight). Data collected on test days one and four were used to assess the influence of practice across days on the selected kinematic and EMG spike parameters. Data collected on test day eight were compared to data collected on test day four in order to assess the reliability of kinematic and EMG spike parameters post-practice. The influence of practice across trials and the influence of gender and inertial load on the selected kinematic and EMG spike parameters were assessed with data collected on all three test days (test days one, four, and eight). Interrelationships

between parameters were assessed with data collected on test day eight.

Results

The findings of the present study demonstrate that it is possible to reliably measure EMG spike parameters during ballistic muscular contractions. The following parameters were measured reliably in this study: maximum speed forearm flexion movement time, maximum speed forearm flexion percent acceleration time; number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike, and spike frequency for biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals. The following parameters were not reliably measured: number of spikes, mean spike amplitude, mean spike duration, mean number of peaks per spike and spike frequency for second biceps brachii burst and triceps brachii burst digital raw EMG signals; and mean spike slope for biceps brachii motor time, end of the first biceps brachii burst, second biceps brachii burst, and triceps brachii burst digital raw EMG signals. In general, EMG spike parameters measured for second biceps brachii burst and triceps brachii burst digital raw EMG signals were not reliable. Mean spike slope was not measured

reliably for any of the four digital raw EMG signals analyzed in this study. A more detailed examination of the factors influencing the reliability of the parameters measured in this study can be found in the section on reliability.

A significant five percent decrease in maximum speed forearm flexion movement time was observed between test days one and four as the result of practice. Subsequently, an additional five percent drop in maximum speed forearm flexion movement time occurred between test days four and eight resulting in a total ten percent decrease in maximum speed forearm flexion movement time over all eight practice days. A short term (across trials) seven percent decrease in maximum speed forearm flexion movement time was found to occur with practice between trials one and ten. Decreases in maximum speed forearm flexion movement time occurred to their greatest extent during the first few trials on test day one. Maximum speed forearm flexion percent acceleration time was stable over trials and test days. The combination of stability of maximum speed forearm flexion percent acceleration time over trials and test days with a concurrent decrease in maximum speed forearm flexion movement time over trials and test days indicates that the magnitude of acceleration of the forearm increased during practice.

The long-term influence of daily practice on EMG spike parameters for biceps brachii motor time digital raw EMG signals was manifested as a twelve percent decrease in mean spike

duration and an eight percent increase in spike frequency between test days one and four. Mean spike amplitude and mean number of peaks per spike for biceps brachii motor time digital raw EMG signals were not influenced by daily practice. Females demonstrated a twelve percent decrease in number of spikes while males demonstrated a eight percent increase in number of spikes for biceps brachii motor time digital raw EMG signals over test days one and four.

No short term influence of practice over trials for biceps brachii motor time digital raw EMG signals on EMG spike parameters was observed. Mechanisms other than alterations in EMG spike parameters such as decreases in biceps brachii motor time are most likely involved in trial to trial decreases in maximum speed forearm flexion movement time.

The influence of daily practice on EMG spike parameters for end of the first biceps brachii burst digital raw EMG signals occurred over all eight practice days rather than over the first four practice days as was the case for biceps brachii motor time digital raw EMG signals. A thirteen percent decrease in mean spike duration and an eleven percent increase in spike frequency was observed over test days one to eight for end of the first biceps brachii burst digital raw EMG signals. No changes over test days were observed for number of spikes, mean spike amplitude, and mean number of peaks per spike for end of the first biceps brachii burst digital raw EMG signals. Across

trials, practice had no influence on EMG spike parameters for end of the first biceps brachii burst digital raw EMG signals.

Males and females performed the maximum speed forearm flexion task differently as evidenced by differences in kinematic parameters between females and males. Maximum speed forearm flexion movement time was twenty three percent slower for females compared to males. Maximum speed forearm flexion percent acceleration time was twelve percent greater for males than females. One EMG spike parameter was found to differentiate females and males. The number of spikes for end of the first biceps brachii burst digital raw EMG signals was thirty six percent greater for females than males. No other EMG spike parameters were influenced by gender for biceps brachii motor time or end of the first biceps brachii burst digital raw EMG signals.

Imposing inertial loads during maximum speed forearm flexions resulted in observable changes in both kinematic and EMG spike parameters. Maximum speed forearm flexion movement time was found to increase as inertial load increased. Load one increased maximum speed forearm flexion movement time seven percent over load zero levels and load two increased maximum speed forearm flexion movement time sixteen percent over load one levels. Maximum speed forearm flexion percent acceleration time was found to be similar for load zero and load one but decreased under load two. Thus, maximum speed forearm flexion

percent acceleration time decreased with increasing inertial load but only under heavy inertial loads. For biceps brachii motor time digital raw EMG signals, load one increased number of spikes seven percent above load zero levels and load two increased number of spikes 10% over load one levels. Imposed inertial load had no other effects on EMG spike parameters for biceps brachii motor time digital raw EMG signals; however, for end of the first biceps brachii burst digital raw EMG signals, changes in mean spike duration and spike frequency were observed in addition to changes in number of spikes. Similar to biceps brachii motor time digital raw EMG signals, load one increased number of spikes three percent above load zero levels and load two increased number of spikes fifteen percent above load one levels for end of the first biceps brachii burst digital raw EMG signals. Thus, the heaviest inertial load (load 2) had the most potent increasing effect on number of spikes. For end of the first biceps brachii burst digital raw EMG signals, load two had the added effect of increasing spike frequency about three percent over load one and load zero levels and decreasing mean spike duration about five percent below load zero and load one levels.

Each EMG spike parameter for biceps brachii motor time digital raw EMG signals positively correlated with the same EMG spike parameter for end of the first biceps brachii burst digital raw EMG signals suggesting that these two signals are characteristically similar. Maximum speed forearm flexion

movement time postively correlated with number of spikes for end of the first biceps brachii burst digital raw EMG signals and negatively correlated with maximum speed forearm flexion percent acceleration time but did not correlate with any other EMG spike parameters. Furthermore, maximum speed forearm flexion percent acceleration time was negatively correlated with number of spikes for both biceps brachii motor time and end of the first biceps brachii burst digital raw EMG signals.

Recommendations

The results of the present study suggest a number of different avenues for further research. An effort to reliably measure EMG spike parameters from triceps brachii burst digital raw EMG signals would provide additional insight into the alterations of EMG signal characteristics in the primary antagonist muscle involved in maximum speed of forearm flexion. To accomplish this objective it is recommended that digital raw EMG signals sampled from the triceps brachii burst be sampled relative to an end of movement event marker and not relative to the beginning and end of EMG activity. Futhermore, it is recommended that the total number of digital raw EMG signals sampled from the triceps brachii burst be increased by increasing

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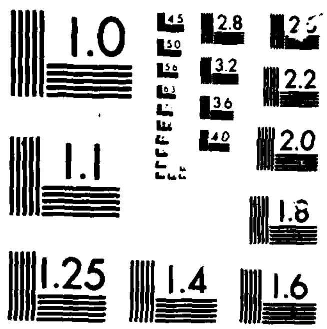

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the number of movement trials performed to insure a reduction in the generally high trial error variance associated with EMG spike parameters measured for the triceps brachii burst in this study.

The application of raw EMG signal analysis techniques to the study of the effect of functional electrical stimulation on speed of human movement is a new area ripe for study. A detailed understanding of the changes in raw EMG signal characteristics during functional electrical stimulation would offer more information than could be gained from only integrated EMG analysis techniques. Moreover, the inclusion of movement impaired individuals in such studies could offer additional understanding of the alterations of central motor commands to the muscle following functional electrical stumulation in such individuals.

The use of additional EMG analysis techniques is another suggestion for future studies that could broaden current knowledge. For example, fast-fourier transformation is a promising method of signal analysis, commonly employed by engineers, recently being applied to raw EMG signals. This technique produces a power frequency spectrum which offers information regarding the distribution of frequency components of a raw EMG signal. The use of new and creative signal analysis techniques in EMG research can only serve to further elucidate electromyography in general.

It is further recommended that a future study be conducted

investigating the influence of age on EMG signal characteristics during improvements in maximum speed of human movement. Studies have already demonstrated that short duration motor unit action potentials are characteristic of older versus younger populations. The importance of such findings to maximum speed of human movement has never been assessed.

The results of this study and other studies have suggested that the mechanisms involved in controlling class A (artificially arrested) and class B (voluntarily arrested) movements are very different. An investigation into maximum speed of human movement in which class A and class B movements are compared is therefore, highly recommended.

The results of the present study reflect alterations in EMG signal characteristics during practice of a maximum speed forearm flexion task in a non-fatigued state. In general, the influence of a fatiguing exercise regimen on maximum speed of human movement and on raw EMG signal characteristics could shed more light on the mechanisms that affect maximum speed.

Conclusions

Based on the results of the present study, the following conclusions are appropriate:

1. Maximum speed of human movement improves under the influence of trial to trial practice and day to day practice.

2. It is possible to reliably measure raw EMG spike parameters during ballistic muscular contractions. The manner in which digital raw EMG signals are sampled may affect dramatically the reliability of subsequent measurements of raw EMG spike parameters.

3. Observable alterations occur in raw EMG spike parameters for biceps brachii motor time digital raw EMG signals during daily practice of a maximum speed forearm flexion task. The influence of daily practice is manifested primarily in terms of mean spike duration decreases and spike frequency increases.

4. Firing rate modulation may be a prominent mechanism involved in alterations in motor unit activation patterning for biceps brachii motor time during daily practice of a maximum speed forearm flexion task.

5. Observable alterations occur in raw EMG spike parameters for end of the first biceps brachii burst digital raw EMG signals during daily practice of a maximum speed forearm flexion task. However, these changes occur slowly. The influence of daily practice is in terms of mean spike duration decreases and spike frequency increases which

require many days to occur. Other mechanisms, such as temporal decreases in the end of the first biceps brachii burst, may therefore be more active than raw EMG spike changes during daily practice.

6. No alterations occur in raw EMG spike parameters during trial to trial practice of a maximum speed of forearm flexion task. Therefore, mechanisms other than alterations in raw EMG spike parameters such as temporal decreases in biceps brachii motor time may more closely characterize EMG changes during trial to trial practice.

7. Gender influences raw EMG spike parameters as well as maximal attainable speed during maximum speed human movements.

8. Resistance to movement influences raw EMG spike parameters as well as maximal attainable speed during maximum speed human movements.

9. Since raw EMG signal characteristics of biceps braachii motor time and end of the first biceps brachii burst seem for the most part unrelated to the ability to improve maximum speed of movement, other neuromotor coordination mechanisms such as agonist-antagonist latency are probably more useful than raw EMG spike parameters in predicting maximum speed of human movement.

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APPENDIX B

Radius of Gyration Formula

Mass (M1)=(Body Weight (kg) x 2.2%)/1.14

Kp*=82.7% x distance from elbow to wrist

MoI**=(M1)(Kp)^2

Kp_1 (Load 1)=[(2.1)(MoI)/M1+M2]$^{-2}$

Kp_2 (Load 2)=[(6.1)(MoI)/M1+M3]$^{-2}$

M2=.45kg lead weight

M3=.90kg lead weight

*radius of gyration

**moment of inertia

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